GENDER DIFFERENCES AND PREDISPOSITION TOWARD A MECHANISM OF ANTERIOR CRUCIATE LIGAMENT INJURY: A BIOMECHANICAL AND ANTHROPOLOGICAL APPROACH

By

MARK DOUGLAS TILLMAN

A DISSERTATION PRESENTED TO THE GRADUATE SCHOOL
OF THE UNIVERSITY OF FLORIDA IN PARTIAL FULFILLMENT
OF THE REQUIREMENTS FOR THE DEGREE OF
DOCTOR OF PHILOSOPHY

UNIVERSITY OF FLORIDA

AUGUST 1999

ACKNOWLEDGEMENTS

Many people have earned my gratitude for their guidance and support during my doctoral education and the completion of my dissertation. First, I would like to thank my wife, Susan, for the love and patience she has provided over the past four years. The completion of my formal education is our shared achievement. I would also like to express gratitude to my parents, Cliff and Margie Tillman. You have provided me with the work ethic, values, and encouragement necessary to achieve my goals.

I would like to express thanks to my committee members. Dr. James Cauruagh has greatly influenced my doctoral career. He was always willing to provide insightful guidance and has acted as an academic example for me to follow as a researcher and a teacher. Dr. Jeff Bauer has also had a great impact on my academic career. With his guidance, I have been able to take advantage many academic opportunities. He was instrumental in the development of my dissertation topic and the choice of technologies to address the issue. Dr. Mark Trimble provided the clinical expertise and guidance necessary for the completion of each stage of my dissertation. Dr. Ed Walsh was my mentor as a master's student and continued to provide me with the relevant technical knowledge and problem solving skills to complete my education. Each of you has taught me lessons that will not be forgotten.

This study would not have been possible without the help of several other individuals. Andi Chiumento laid the groundwork for this study by successfully completing her thesis. Dr. Tom Kaminski provided clinical expertise in the developmental stage of this project. Kendra Smith and Dr. Anthony Falsetti supplied anthropological knowledge crucial to this project. Dr. John Harris assisted with the creation of computer code for data analysis. Chris Hass provided support and assistance in my greatest times of need. Finally, I am forever grateful to Lynn Pattishall. She helped collect and analyze data in every stage of my dissertation and also provided emotional support during the less pleasant stages of the project.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

Abstract of Dissertation Presented to the Graduate School
of the University of Florida in Partial Fulfillment of
the Requirements for the Degree of Doctor of Philosophy

GENDER DIFFERENCES AND PREDISPOSITION TOWARD
A MECHANISM OF ANTERIOR CRUCIATE LIGAMENT INJURY:
A BIOMECHANICAL AND ANTHROPOLOGICAL APPROACH

By

Mark Douglas Tillman

August 1999

Chairman: James H. Cauraugh
Major Department: Exercise and Sport Sciences

Female participation in athletics has increased significantly over the years since Title IX. Epidemiological studies have reported that females suffer noncontact anterior cruciate ligament (ACL) injuries at higher rates than males. This difference in ACL injury rates has been scrutinized, but no definitive relationships have been identified between musculoskeletal alignment and ACL injury. This three study project investigated anatomy possibly related to ACL injury (Notch width index, notch area index, notch shape index,quadriceps angle, thigh foot angle, subtalar joint ratio), a common mechanism of ACL injury (tibiofemoral rotation), and a possible intervention (in-shoe orthotics).

Two hundred (100 male and 100 female) cadaver femurs from the Terry Collection (Smithsonian) were analyzed to evaluate intercondylar notch geometry

between genders in Study I. One measure, notch shape index, identified a larger index for males in comparison to females.

Study II measured the musculoskeletal alignment of 57 individuals (30 males and 27 females). The analysis revealed a larger quadriceps-angle for females than males. Two additional measures, thigh foot angle and subtalar joint ratio, did not vary between genders.

In Study III, 51 participants (28 males and 23 females) were evaluated relative to tibiofemoral rotation while running and landing from a jump. In-shoe orthotics were introduced, and a prediction equation for maximum internal tibiofemoral rotation was computed. Neutral orthotics increased tibiofemoral rotation compared to the medial orthotics.

The landing task analysis revealed two significant interactions (Posting x Trial x Gender; Posting x Gender). Analysis of the third trial indicated that tibiofemoral rotation for males in the no orthotic condition was greater than tibiofemoral rotation for females in the neutral posting condition. No main effects for tibiofemoral rotation or orthotic posting were found between genders. A stepwise linear multiple indicated activity was the only predictive variable.

The differences in notch shape index and quadriceps-angle may partially explain the discrepancy in ACL injury rates between genders. The findings further suggest that gender has minimal influence on tibiofemoral rotation, that tibiofemoral rotation is relatively unaltered by orthotic intervention, and that jumping sports are associated with a higher amount of tibiofemoral rotation than running sports.

CHAPTER 1
GENDER DIFFERENCES AND ACL INJURIES

Introduction

Congress passed Title IX in 1972 as part of the Education Amendments Act, and it requires that colleges and high schools that receive federal funding do not discriminate on the basis of gender regarding any educational activity. Athletic participation falls under this jurisdiction. Title IX has led to a steady increase in athletic participation for females, and the number of injuries to female athletes has grown concurrently.

Overall, male and female athletes performing in similar sports are injured at similar rates. However, an important exception to this rule is that female athletes suffer injuries to the anterior cruciate ligament at a higher frequency than their male counterparts (Huston & Wojtys, 1996). Although the cause of disparity in injury rate is not fully understood, numerous extrinsic and intrinsic factors are believed to be associated with ACL injury. Despite extensive research, no strong relationship between any single contributing factor and ACL injuries has been established.

The current project involved a three-phase multidisciplinary effort to study anatomy believed to be related to ACL injury and the kinematics of a common mechanism of ACL injury. The work was a simultaneous evaluation of specific anatomic variations in male and female lower limb morphology and their relationship to movements that are known to be mechanisms of non-contact ACL injury. The project consists of three separate studies.

Study I included statistical analysis of cadaveric skeletal anatomy to evaluate data describing morphological similarities and differences for each sex. Specifically, the geometry of the distal end of the femur was analyzed with digital images. Study II consisted of anthropometric analysis of anatomical variants in living subjects to relate skeletal variation to living individuals. Geometry of the lower extremity suspected of predisposing individuals to ACL injury was compared between college-aged males and females. Study III involved a biomechanical analysis, as the effects of morphology on movement (i.e., tibiofemoral rotation) were determined. A prediction equation between anatomical profile and movement characteristics was calculated to determine if certain anatomical profiles are associated with tibiofemoral rotation. Additionally, an intervention was introduced to determine whether tibiofemoral rotation can be controlled with in-shoe orthotics.

The three studies were expected to reveal the following: (a) anthropometric differences can be confirmed to exist between males and females via post mortem skeletal analysis and measures on living individuals for specific parameters believed to be related to ACL injury, (b) thigh foot angle and tibiofemoral rotation are related, (c) calcaneal inversion/eversion and tibiofemoral rotation are related, (d) orthotic posting can limit excessive internal tibiofemoral rotation, and (e) clinical methods can be developed to identify athletes predisposed to a mechanism of ACL injury.

This project presented a unique approach to the pathomechanics of traumatic injury of the lower limb that integrated several disciplines. The results are important for physicians, therapists, trainers, coaches, military, and exercise professionals who will be able to help design programs and interventions to lower injury rates among individuals who exhibit high-risk morphology. Female as well as male athletes will benefit from this effort.

Review of Literature

General Introduction

The anterior cruciate ligament (ACL) is considered the most important ligament of the knee (Markolf et al., 1995). It prevents anterior translation of the tibia relative to the femur and also acts as a rotational stabilizer. ACL injuries are a common source of disability in the United States (Goris & Graf, 1996). Injury to the ACL may begin a course of events to include instability, further ligamentous injury, meniscal injury, and arthritis (Markolf et al., 1995). Advances in surgical techniques and rehabilitation permit many individuals to return to their pre-injury state of activity. However, some athletes never regain their pre-injury level of performance and may be prevented from further participation. No ideal substitute exists for an athlete's normal ACL (LaPrade & Burnett, 1994).

Increased participation of females in collegiate sports since the passing of the Title IX Educational Assistance Act of 1972, requiring federally-funded institutions to provide equal opportunity to females in all areas, including athletics, has been accompanied by an increased incidence of athletic injuries in females (Arendt & Dick, 1995; Hutchinson & Ireland, 1995). Overall, the frequency of athletic injuries among female athletes is nearly equal to those among males for a given sport with similar rules. However, females are substantially more susceptible than males to acute noncontact injury of the ACL (Loudin, Jenkins, & Loudin, 1996; Arendt & Dick, 1995; Hutchinson & Ireland, 1995; Arendt, 1994). Although the cause of disparity in injury rate is not fully understood, numerous extrinsic and intrinsic factors are believed to be associated with ACL injury. Despite extensive research, no strong relationship between any single contributing factor and ACL injury has been established.

Noncontact ACL injuries most often occur during knee hyperextension and forced internal tibiofemoral rotation (Loudin et al., 1996). These movements happen during closed-chain activities such as landing from a jump, quick changes of direction during running, or sudden deceleration (Nawocezenski, Cook, & Saltzman, 1995; Gray et al., 1985; Griffis, Vequist, & Yearout, 1989). During a closed chain activity, the movement of one joint influences all of the joints in the chain (Inman, 1981; Inman, 1976). This dictates the need for an analysis of the entire lower extremity, including, static anthropometric measures and dynamic biomechanical measures.

Analyzing the entire kinematic chain of the lower extremity, in combination with anthropometric measurements that give insight into the shape and function of the whole body may identify morphologies or combinations of morphologies significantly related to ACL injury risk. Knowledge of predisposing morphologies will provide a crucial foundation for the design and implementation of effective measures for injury prevention.

Knee Anatomy

The knee is the largest and one of the most complex joints in the body (Tortora, 1999). The knee joint is composed of the tibiofemoral and patellofemoral joints. The tibiofemoral joint is the articulation between the condyles on the distal end of the femur and condyles on the proximal end of the tibia. The patellofemoral joint is the articulation between the distal end of the femur and the patella.

The geometry of the tibiofemoral joint provides little stability (Irrgang, Safran, & Fu, 1996). The femoral condyles are convex in both the anterior-posterior and medial-lateral directions.The medial condyle of the tibial plateau is concave in the anterior-posterior and medial-lateral directions. The lateral condyle of the tibial plateau is concave in the medial-lateral direction, but is convex in the anterior-posterior direction. The menisci lie between the femoral and tibial condyles. The menisci function to provide stability by increasing the concavity of the tibial plateau, as well as, to increase the contact area between the femur and tibia.

The knee is the joint most vulnerable to injury because of the high stresses to which it is subjected and its stability is supplied solely by ligaments and tendons (Tortora, 1999). The stability of the knee depends on static and dynamic restraints. Static constraints are non-contractile in nature and include menisci, the joint capsule, and ligaments. Dynamic constraints are the muscles that cross the knee joint and include the quadriceps, hamstrings, and gastrocnemius.

The primary ligamentous restraints of the knee are the collateral and cruciate ligaments. The medial collateral ligament (MCL) attaches on the medial epicondyle of the femur and the medial side of the tibia just below the tibial plateau. The MCL resists valgus forces that would tend to separate the medial compartment of the tibiofemoral joint. The lateral collateral ligament (LCL) attaches on the lateral epicondyle of the femur and the head of the fibula. The LCL resists varus forces that would tend to separate the lateral compartment of the tibiofemoral joint. The posterior cruciate ligament (PCL) runs anteriorly from the posterior aspect of the tibia to the lateral wall of the medial femoral condyle. The PCL is the main restraint to posterior translation of the tibia relative to the femur. The ACL originates from the anterior part of the intercondylar area of the tibia and extends superiorly, posteriorly, and laterally through the intercondylar notch of the femur to attach to the posterior part of the medial side of the lateral condyle of the femur (Marieb & Mallatt, 1992). In this position, the ACL limits anterior translation of the tibia relative to the femur, as occurs in hyperextension of the knee, and internal tibiofemoral rotation. Internal tibiofemoral rotation is the relative rotation between the tibia and femur that is caused by inertial forces acting on the femur. It is readily accepted that the ACL serves as the primary stabilizer of the knee. The structures of the knee are shown in Figure 1-1.

<u>Injury Epidemiology</u>

ACL injuries are a common source of disability in the United States (Goris & Graf, 1996). Approximately 100,000 ACL injuries occur each year (Feagin et al., 1987). These injuries result in lost time from work, decreased physical participation, and may require surgical reconstruction.

The problem of knee instability resulting from ACL injury and possible solutions were addressed as early as the 1900's. Robson (1903) stated that the crucial (cruciate) ligament may be torn without serious damage to the knee joint and that its successful repair by operation was feasible. Ninety-five years later, as sports participation has increased and surgical techniques have improved, this statement has become true. Approximately 50,000 ACL reconstruction surgeries are performed each year in the United States (Paulos, 1992). Alternatively, 50,000 individuals choose to treat the condition conservatively using physical therapy alone.

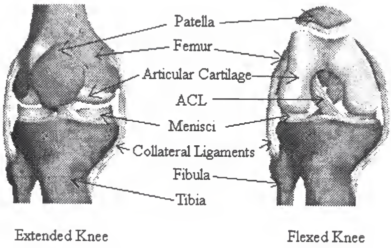


Figure 1-1. Anatomy of the knee in extended and flexed positions.

Physical, mental, and monetary costs are associated with ACL injury and repair. The physical and mental costs are difficult to quantify; however, the monetary costs are not. Individuals choosing to surgically reconstruct their ACL must pay surgical fees as well as rehabilitation costs. The average cost for an outpatient ACL reconstruction is $8,815 (Novak et al., 1996). Thus, approximately $440,750,000 is spent on ACL surgeries in the United States each year. The average cost for a conventional ACL rehabilitation at a physical therapy clinic is $6,930 (Timm, 1997). Therefore, $346,500,000 is spent on rehabilitation for ACL surgeries each year in the United States. The total health care costs associated with ACL repair reach nearly $800 million per year.

Injury Rates

Musculoskeletal injuries are believed to be sport-specific and not gender-specific (Arendt & Dick, 1995). This belief has held true except in the area of knee injuries and specifically ACL injuries. Awareness of medical issues for female athletes has increased as female participation in athletics has increased. Since Title IX was enacted, an alarming trend has been detected. Comparisons of noncontact injuries between athletes of different genders has confirmed that female athletes suffer ACL injuries more often than their male counterparts given the same sport and playing conditions (Arendt & Dick, 1995). A noncontact injury is one in which the athlete does not collide with another athlete or structure besides the ground. Arendt and Dick (1995) reported that collegiate female soccer players were more than twice as likely to sustain an ACL injury as collegiate male players and collegiate female basketball players were four times as likely to sustain an ACL injury as collegiate male players. Malone and colleagues (1993) reported that collegiate female basketball players were eight times more likely to suffer an ACL injury compared to their male counterparts. The National Collegiate Athletic Association Injury Surveillance System supported these findings by releasing data indicating that female athletes injured their ACLs almost eight times as frequently as their male counterparts (Pearl, 1993).

Mechanisms of Injury

Sports involving physical contact among players are expected to produce ACL injuries (Huston & Wojtys, 1996). However, the majority of ACL injuries occur in a noncontact fashion. Noyes and colleagues (1983) reported that 78% of ACL injuries were noncontact and occurred during landing from a jump, quick changes in direction while running, or sudden deceleration. In a survey of ACL injured skiers, 88% of the injuries occurred in the presence of tibiofemoral rotation (Jarvinen et al., 1994). The two primary mechanisms were a combination of valgus knee position and external rotation and a combination of flexion and internal rotation. The predominant injury mechanisms for the female skiers studied were knee flexion and internal rotation (Jarvinen et al., 1994). Hess and colleagues (1994) investigated 151 sports-related ACL injuries and observed a similar trend with the majority of injuries occurring as a result of flexion and internal tibial rotation. Emerson (1993) noted the prevalence of ACL injuries in basketball players as the result of deceleration and change of direction when the tibia is internally or externally rotated. Tibial rotation appears to be the common link among the different mechanisms associated to the injury and therefore, will be examined during the dynamic phase of the present work.

Factors Related to Injury

Many theories have been proposed to explain the discrepancy between rates of injury for males and females. The proposed theories are based on anatomical elements believed to influence knee mechanics. These elements are divided into extrinsic and intrinsic factors. Extrinsic factors include muscular strength, muscular coordination, shoe-surface interface, skill level, and level of conditioning (Arendt & Dick, 1995). Instrinsic factors include joint laxity, limb alignment, notch dimensions, and ligament size (Arendt & Dick, 1995). Extrinsic factors can be controlled directly by the individual, including level of physical fitness, level of flexibility, excessive physical training, type of movement, muscular strength, level of skill and/or conditioning, shoe/surface interface, and cigarette smoking (Cowan et al., 1996; Arendt & Dick, 1995). Intrinsic factors are anatomical variations that are not directly controllable; however, modification of extrinsic factors, such as implementation of tailored strengthening programs, may counteract their effects. The focus of this work lies in the area of identifying intrinsic factors that may predispose individuals to ACL injury by determining their effects on a common injury mechanism.

Cowan and colleagues (1996) determined that high degree of genu valgus and Q-angle are associated with higher risk of overuse injures in male military trainees. The wide pelvis and interacetabular distance necessary for childbirth, combined with the need to maintain the center of mass inside the base of support for effective bipedality, may result in a higher frequency of increased genu valgus and Q-angle in the female population. Likewise, a smaller intercondylar notch width index (the ratio of intercondylar notch width to condylar width), which may occur more frequently in the female population, is thought to impinge upon the attachment sites of the ACL and create a situation in which the risk of shearing or tearing is highly probable (Malone et al., 1993; Souryal & Freeman, 1993; Huse, Himeno, Coventry, & Chao, 1990; Souryal, Moore, & Evans, 1988; Houseworth, Mauro, & Mellon, 1987; Noyes, Matthews, & Butler, 1983). Although the increased frequency of genu valgus, high Q-angle, and narrow intercondylar notch within the female population likely accounts for some proportion of the difference in injury risk between males and females, the relationship between biomechanics of the female lower limb musculoskeletal morphology and rate of lower limb injury has not been well documented or explicitly tested.

Skeletal Morphology. Morphological variations of the lower extremity, including degree of limb alignment, intercondylar notch width index, degree of lateral displacement of the femur over the knee, excess navicular drop, excess subtalar joint pronation, and rotational alignment of the lower extremity are among the intrinsic factors believed to contribute to risk of ACL injury (Heugel et al., 1999; Loudin et al., 1996; Cowan et al., 1996; Arendt & Dick, 1995; Nawocezenski, Cook, & Saltzman, 1995). Small differences in morphology at different parts of the kinematic chain may interact synergistically to significantly increase injury risk. For example, internal rotation of the tibia, a mechanism of ACL injury, is required when the talus moves into adduction upon subtalar joint pronation due to the congruence of the talus with the ankle mortise (Inman, 1981; Inman 1976). Whether an injury occurs, however, may depend to some degree on morphology above the knee. For example, a large Q-angle may pull the patella laterally, making the knee more unstable and thus, increase the risk of ACL damage due to internal tibial rotation.

Skeletal anatomy of the pelvis and thigh is known to differ between the sexes (Hutchinson & Ireland, 1995). Generally, male pelvic and lower limb anatomy represents optimization for bipedality. Characteristic female anatomy is the result of optimization for both bipedality and childbirth (Rosenberg, 1992; Weaver, 1980). The additional musculoskeletal demands placed on females for childbirth may create a predisposition to ACL injuries. The mechanics of human bipedality require that the pelvis be widened and tilted relative to quadrupeds and that the knees and feet allow for positioning such that the center of mass of the body remains within the base of support during gait. Angulation of the femur, resulting in a degree of genu valgus, is an adaptation that helps to accomplish this. It is defined as a ratio greater than 1.0 of pelvic width (distance between ASIS) to patellar width (distance between centers of patellae) in standard anatomical position and is associated with bipedality (Heiple & Lovejoy, 1971; Lovejoy & Heiple, 1970). Angulation of the femur is graphically depicted in Figure 1-2.

Expansion of the pelvic inlet and outlet, necessary for the passage of the relatively large head of the human fetus during childbirth, results in a broader ilium, elongated supra-pubic ramus, wide sciatic notch, oblique sub-pubic angle, and large interacetabular distance relative to the male counterparts (Bass, 1987; Krogman, 1968). Additionally, females generally exhibit less robusticity (thickness and strength), smaller joint surfaces, and possibly a smaller intercondylar notch width index compared to males (Hsu et al., 1990; Stewart, 1979).

Morphology of the pelvis and thigh differs between males and females (Nawocezenski, Cook, & Saltzman, 1995, Malone, Hardaker, & Garrett, 1993). Conversely, morphology of the lower leg and foot is not known to differ between the sexes and has recently been implicated as a risk factor for musculoskeletal injury of the knee. Lower limb morphology may contribute to the disparity in non-contact ACL injury

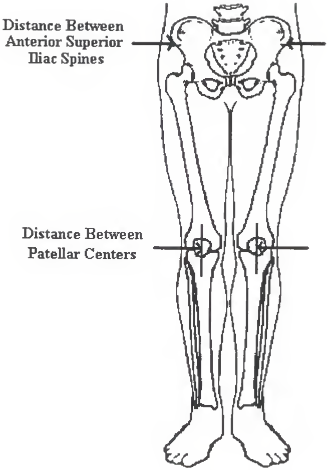


Figure 1-2. Graphical depiction of pelvic width and patellar width while standing in standard anatomical position.

rate by differential interaction with sexually dimorphic characteristics above the knee. Loudin et al. (1996), used regression analysis to conclude that the combination of knee recurvatum, excess navicular drop, and excess subtalar joint pronation is a strong discriminator between females with ACL injury and noninjured individuals. Additionally, a relationship between the rotational alignment of the lower extremity and ACL tears in female athletes has been identified (Huegel et al., 1998). The critical variable, as reported by Heugel and colleagues (1999), in determining risk of injury is external tibial torsion and is measured as thigh foot angle (TFA).

Intrinsic Factors

Intercondylar Notch Geometry. Stenotic (narrow) intercondylar notches have been implicated as a possible risk factor for ACL injuries. However, there is considerable debate about notch width varying between injured and uninjured individuals. Additionally, controversy exists concerning notch width variance between males and females. Notch width is most often reported as a Notch Width Index (NWI). The NWI is calculated as the ratio of intercondylar notch width to the width of the femoral condyles (Muneta, Takakuda, & Yamamoto, 1997; Goris & Graf, 1996; LaPrade & Burnett, 1994). The procedure for determining NWI is shown in Figure 1-3. Values below 0.2 are considered stenotic (excessively narrow).

Anderson and colleagues (1987) detected signficant differences between ACL injured individuals and normal individuals when NWI was used as a basis of comparison. Souryal and Freeman (1993) deduced that athletes with stenotic intercondylar notches were more likely to suffer ACL injuries. LaPrade and Burnett (1994) studied 213 Division I collegiate athletes and concluded that intercondylar notch stenosis is associated with

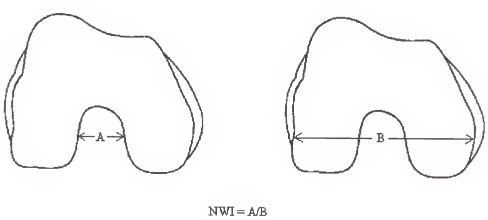


Figure 1-3. NWI is determined by dividing the width of the intercondylar notch (A) by the width of the femoral condyles (B).

ACL injury, but no differences were detected between the sex of the athlete and NWI. Teitz, Lind, and Sacks (1997) compared individuals that had suffered noncontact ACL injuries with normal individuals and detected no differences between notch width indices for males and females or between normal healthy individuals and injured individuals. These authors also noted that a considerable amount of overlap exists for the ranges of NWI for males (0.19-0.32) and females (0.16-0.32).

Other researchers have used a different approach to analyzing the intercondylar notch. Houseworth and colleagues (1987) calculated the area of the intercondylar notch from roentgenograms of injured and uninjured individuals. The authors concluded that a narrow posterior arch of the intercondylar notch may be a risk factor related to ACL injury. Another interesting study dealt with the cross-sectional area of the anterior cruciate ligament and the NWI. Muneta, Takakuda, and Yamamoto (1997) examined 16 cadaver knees and concluded that the width of the intercondylar notch was not related to the size of the ACL. Two schools of thought exist concerning intercondylar notch stenosis. The first believes that a narrow notch will have a correspondingly smaller ACL. A smaller ACL would be more likely to rupture. The second hypothesizes that a narrow notch surrounding a normal size ACL would provide insufficient space for the ligament to function normally. The reduced space could pinch the ACL and cause damage. The work of Muneta, Takakuda, and Yamamoto (1997) supports the second theory as a mechanism of ACL rupture. Another potentially critical aspect of the intercondylar notch is the shape of the notch itself. Intercondylar notch shapes may be classified as inverted U- or A-shaped (Hutchinson & Ireland, 1995). Anderson et al. (1987) also observed the inverted

U-shape for normal notches, but observed narrow notches to be more waveshaped and noted the need for additional research in the area of intercondylar notch shape.

Q-angle. Another anatomical factor that has been implicated as predisposing to ACL injuries is Q-angle. The Q-angle is defined as the acute angle formed by a line drawn from the center of the tibial tuberosity through the midpoint of the patella and a line running from the midpoint of the patella to the anterior superior iliac spine, as shown in Figure 1-4 (Schulthies, Francis, Fisher, & Van De Graaff, 1995). The Q-angle is measured with the knee in full extension and is intended to provide an estimate of the direction of the resultant force produced by the quadriceps (Schulthies et al., 1995; Terry, 1980). Schulthies et al. (1995) concluded that a clinically measured Q-angle is representative of the angle of pull of the quadriceps muscle, but is significantly lower than the actual value of the angle of pull. Therefore, Q-angle measures will underestimate the lateral force applied to the patella by the quadriceps muscle.

The majority of the Q-angle research has indicated that females have higher Q-angles than their male counterparts (Hsu et al., 1990; Woodlang L. H. & Francis, R. S., 1992). Reports indicate that the Q-angle for females range from 2.0° to 8.5° greater than males (Agletti, Insall, & Cerulli, 1983; Hvid, Andersen, & Schmidt, 1981). Conversely, Livingston and Mandigo (1997) detected no significant differences in Q-angle between males (10.5°) and females (12.2°) for a group of fifty young individuals with no history of knee disorder. Additionally, no research has been reported in which the mean Q-angle in males exceeded that for females (Livingston, 1998).

Large Q-angles have been associated with lower limb injuries and knee extensor dysfunction (Cowan et al., 1996; Ciullo, 1993; Hunter, 1984). Theoretically, a greater Q-

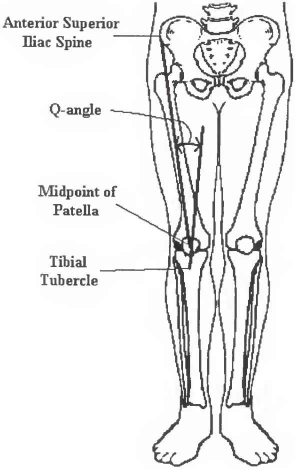


Figure 1-4. The Q-angle is defined as the angle formed between the line drawn between the center of the patella and the anterior superior iliac spine (ASIS) and from the tibial tuberosity through the center of the patella.

angle increases the horizontal component of the quadriceps muscle force and pulls the patella laterally. Increased horizontal force on the patella exposes the ligaments and menisci of the knee to greater loads and renders these structures vulnerable to injury (Hsu et al., 1990).

Thigh Foot Angle. Thigh foot angle (TFA) has recently been linked with injuries to the lower limb and specifically ACL injuries (Heugel et al., 1999). TFA is a combined measure of tibiofibular torsion and alignment of the structures distal to the talocrural joint (Stuberg, Temme, Kaplan, Clarke, & Fuchs, 1991). TFA is measured by positioning the individual in a prone position, flexing the knee to 90°, moving the ankle to a neutral position, and determining the angle formed by the intersection of a line drawn from the midpoint of the calcaneus and the second toe and the long axis of the femur (Cheng, Chan, Chiang, & Hui, 1991).

Excessive internal tibiofemoral rotation may also result from excessive external tibial torsion, measured as TFA. It appears that high TFA induces an external torque during footstrike. This external torque would have to disrupt the medial structures of the knee to affect the ACL, but Heugel et al. (1999) did not observe this phenomenon. Rather, they suggest that an increase in TFA may cause poor foot plant in a position of increased external rotation relative to the knee. Tiberio (1987) noted that external tibial torsion should be considered an indirect factor that leads to excessive pronation of the subtalar joint. A possibility exists that, as the foot lands in external rotation, the forward momentum of the leg may force the subtalar joint into pronation as the center of mass of the entire body moves anteriorly over the foot.

Subtalar Joint Axis. The bony geometry of the foot and ankle has been implicated as a risk factor for injuries to the lower limb. Specifically, excessive foot eversion and abnormal tibial rotation have been associated with knee injuries (Hintermann & Nigg, 1993). The relationship between these two movements is determined by the subtalar joint axis. The subtalar joint is defined as the articulation between the talus superiorly and the calcaneus and navicular inferiorly (Rockar, 1995). The subtalar joint axis passes through the talus and extends from below on the lateral side of the heel, upward, forward, and medially (Inman, 1976; Hicks, 1953; Manter, 1941). Manter (1941) determined the location of the axis on cadaver lower limbs. The mean axis lies 42° above the horizontal plane and 16° medial to the midline of the foot (see Figure 1-5). However, great variation in the location of the axis exists among different individuals. A lower axis will result in greater frontal plane movement which will occur as calcaneal eversion and inversion. A higher axis will result in greater horizontal plane movement, which will occur as internal and external tibial rotation. Therefore, a lower axis accompanied by greater eversion of the foot may predispose an individual to greater injuries of the foot, while a higher axis with greater tibial rotation may predispose an individual to more knee injuries (McClay & Bray, 1996).

The relationship between tibial rotation (a transverse plane motion) and inversion/eversion of the subtalar joint (a frontal plane motion) is not fixed (Root, Orien, & Weed, 1977). Depending on the height of the subtalar joint axis, some individuals may have proportionally more internal tibial rotation with a fixed amount of eversion than others. Subtalar joint pronation is a triplanar motion with motion occurring around a single axis, which is oblique to the three cardinal planes. As with any planar motion, the

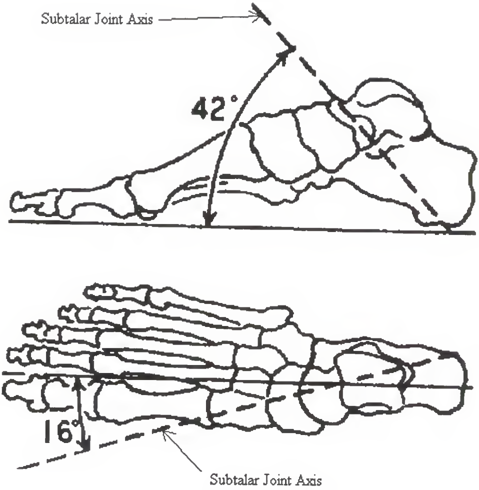


Figure 1-5. Diagram showing the location of the average subtalar joint axis (depicted by the dashed line). From Manter (1941).

movement occurs in the plane that is perpendicular to the joint axis. Normally, the subtalar joint axis lies closest to the sagittal plane, thus, the majority of motion occurs in the frontal and transverse planes. A high subtalar joint axis lies closer to the frontal plane and limits movement in that plane. Therefore, all compensatory motion must occur in the transverse plane as rotation. Individuals with high subtalar joint axes may be more at risk of ACL injury upon compensatory pronation of the subtalar joint than those with lower subtalar joint axes because they have a greater amount of internal tibial rotation with a given amount of subtalar joint pronation (see Figure 1-6).

Subtalar joint pronation, a combination eversion, abduction, and dorsiflexion, compensates for a variety of lower extremity malalignments (Kitaoka, Lundberg, Luo, & An, 1995). For example, when forefoot and/or calcaneal varus deformities are present, subtalar joint pronation allows the plantar surface of the foot to contact the ground appropriately during closed-chain activities. The internal rotation of the tibia occurring concurrently with subtalar joint pronation affects knee kinematics. If the magnitude of internal tibial rotation is proportional to the degree of subtalar joint pronation, excessive subtalar joint pronation may translate into excessive internal tibiofemoral rotation, increasing the distance between the attachment points of the ACL and inducing tension in the ligament (Ahmed, Burke, Duncan, & Chan, 1992). Therefore, excessive internal rotation of the tibiofemoral joint may predispose the ACL to injury.

The close relationship between tibial rotation, subtalar joint pronation, and malalignments of the foot suggests that management of the compensatory motions through foot orthotic intervention may control excessive motions and reduce the risk of knee injury. Foot orthotics are commonly used to control subtalar joint movement and are

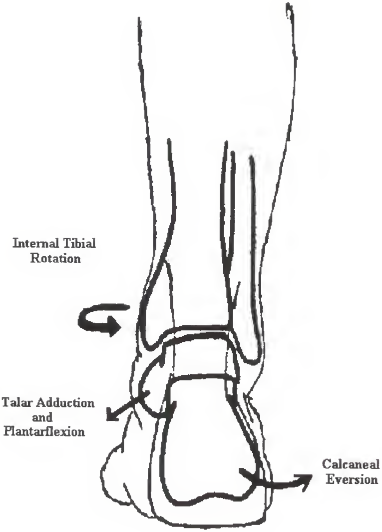


Figure 1-6. Diagram depicting closed chain pronation and concurrent internal tibial rotation. From Tiberio (1987).

intended to restore dynamic stability and reduce compensatory pronation of the subtalar joint during the initial stance phase of gait (McCulloch, Brunt, & Vander Linden, 1993).

Chiumento (1998) conducted an initial attempt to identify the link between tibiofemoral rotation and orthotic interventions for female participants. Analyses were performed for walking and jumping tasks and data indicated that orthotic interventions influence tibiofemoral rotation when landing from a jump. Although the study was limited by the number of participants ($\underline{n} = 7$) in the jumping portion, it was the first to relate in-shoe orthotics to a common mechanism of ACL injury. These results, in combination with the body of literature related to ACL injury, have identified the need for an in-depth analysis for males and females of intrinsic factors suspected of predisposing individuals to injury, a common mechanism of injury, and the effect of in-shoe orthotics on transverse plane kinematics of the knee during dynamic activities.

Completed Research: Three Studies

A three-phase effort was completed in the current project to achieve the overall goals of (a) detecting the existence of specific anatomic variations in male and female lower limb morphology and (b) determining if these anatomic variations are related to a common mechanism of noncontact ACL injury. Additionally, an intervention (three types of orthotic posting) was tested in an attempt to influence rotation of the knee during dynamic situations.

Study I. The first study of this project involved an anthropometric study of cadaver femur anatomy. The size and shape of the intercondylar notch have been implicated as possible risk factors related to ACL injury. The results of previous research have been inconclusive. Some data indicate that females have a more narrow intercondylar notch than males, while other data show that no difference exists between females and males. Orthopedic surgeons have also observed that the shape of the intercondylar notch varies between females and males. Their subjective observations have not been confirmed quantitatively.

Cadaveric skeletal data from the remains of individuals stored in the Museum of Natural History in the Smithsonian Institution, Washington, D.C. were studied. Fourteen hundred individuals are housed in the Terry Collection in Washington. From this collection, a total of 200 femurs were analyzed. The distal end of 100 female and 100 male femurs was digitally photographed. From these photographs, indices of notch width, notch shape, and notch area were calculated. This study was designed to detect anatomical differences between females and males related to the bony geometry of the intercondylar notch. Confirmation of anatomical differences between males and females may give insight into possible predisposing anatomical factors to ACL injury.

Study II. The second phase focused on anthropometric measures of living individuals. Physically active college-aged adults were recruited to participate. Anthropometric measures associated with ACL injury were measured directly using goniometers and indirectly using digital images. This study was designed to detect differences in musculoskeletal alignment of the lower extremity between males and females. Again, published data are contradictory concerning the existence of musculoskeletal differences between the sexes and possible relationships to ACL injury.

Measures of Q-angle, thigh foot angle, subtalar joint motion, and flexibility were collected on participants. All of these factors are suspected to influence tibial rotation and possibly tibiofemoral rotation. Tibiofemoral rotation is a common mechanism of ACL injury and therefore, may be dependent on the musculoskeletal factors considered here.

Study III. The third phase of this project was a kinematic analysis of knee mechanics during dynamic activities. Specifically, tibiofemoral rotation was analyzed using high-speed video. The same participants from Study II performed a series of jump landings and ran on a treadmill. Landing from a jump and running are closed-chain activities that are associated with some mechanisms of ACL injury. Additionally, in-shoe orthotics were introduced in an attempt to manipulate tibiofemoral rotation during the dynamic activities.

The anthropometric data collected in Study II were combined with the kinematic data of Study III to explore the existence of relationships between static musculoskeletal measures suspected of predisposing an individual to ACL injury and dynamic measures of a mechanism of ACL injury (tibiofemoral rotation). Successful identification of factors predisposing individuals to excessive tibiofemoral rotation will open avenues for longitudinal research. Long-term research may confirm the link between static musculoskeletal measures and ACL injury, as well as, the possibility of introducing in-shoe orthotics as a possible intervention. An outline of the completed studies and directions for future research appears in Figure 1-7.

Summary

Females suffer noncontact ACL injuries at higher rates than their male counterparts. Several intrinsic and extrinsic factors have been analyzed in an attempt to explain the discrepancy in injury rates, but results for any single factor have been inconclusive. One of the more consistent factors related to noncontact ACL injury appears to be a mechanism of injury. Specifically, tibiofemoral rotation is commonly associated with ACL injury. A comprehensive study of intrinsic factors, a common mechanism of injury, and an intervention designed to control knee rotation provides a more complete picture skeletal morphologies and movement patterns linked to ACL injuries and differences in injury rates between males and females. Therefore, a three phase project has been executed to address these issues. The first study of the project was intended to detect skeletal differences between males and females. The second was directed toward detecting musculoskeletal differences between males and females. The third study compared males and females for a mechanism of injury and to develop a prediction equation relating anthropometric measures to a common mechanism of injury. The successful completion of this project could lead to long term injury tracking research to develop a technique for predicting which individuals are most likely to suffer an ACL injury and to develop an intervention to reduce the risk of injury for individuals predisposed to noncontact ACL injury.

The human lower extremity is a complicated mechanism with the knee joint being the most critical link. The integrity of the entire mechanism can become compromised when the primary stabilizer of the knee is injured. The links adjacent to the knee must not be considered insignificant and they may have considerable influence on knee mechanics during closed chain activities. Therefore, an effective evaluation of the lower extremity must include both static (anthropometric) and dynamic (kinematic) analyses. A combination of these data gives an understanding into the morphology or morphologies that may be related to knee injuries, specifically ACL injuries. Additionally, knowledge of these data may allow for the development of injury intervention strategies.

The completed studies present a unique approach to the pathomechanics of traumatic injury of the lower limb that will integrate several disciplines. The success of the project depended on documentation and comparison of the normal range of anatomical variation of males and females related to ACL injury and analysis of a mechanism of ACL of injury associated with the variants. Additionally, the effects of an intervention (in-shoe orthotics) on a mechanism of ACL injury were evaluated. Results may prove important to physicians, therapists, trainers, coaches, military, and exercise professionals who will be able to help design programs and interventions to lower injury rates among individuals, female and male, who exhibit high-risk morphology.

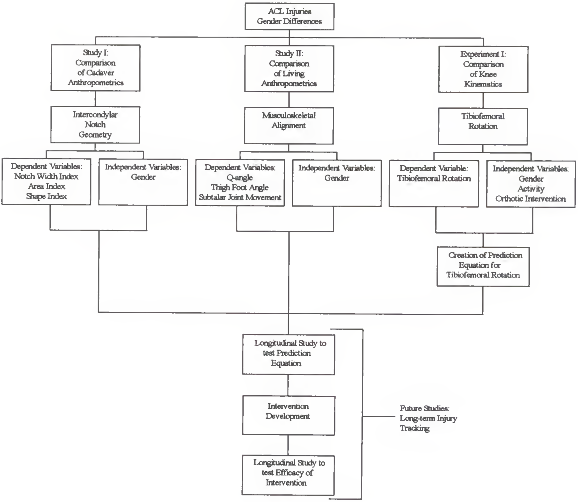


Figure 1-7. Flow chart describing the multiphase effort including Studies I, II, and Study III.

CHAPTER 2
CADAVER ANTHROPOMETRY: DO MALES AND FEMALES DIFFER IN INTERCONDYLAR NOTCH GEOMETRY?

As a first step in determining why female athletes suffer noncontact ACL injuries at higher rates than male athletes, skeletal factors that may differ between genders and may predispose an individual to injury were analyzed. Study I was designed to create a database of lower extremity skeletal anthropometric knowledge that can discern differences between males and females. This data base provided the groundwork for study in the areas of musculoskeletal alignment and mechanisms of ACL injury.

The geometry of the intercondylar notch has been implicated as a possible risk factor related to ACL injury. Some data indicate that females have a more narrow intercondylar notch than males (Shelbourne, Davis, & Klootwyk, 1998), while other data show that no difference exists between females and males (Teitz, Lind, and Sacks, 1997; LaPrade and Burnett, 1994). Orthopedic surgeons have also observed that the shape of the intercondylar notch varies between females and males. However, their subjective commentary has not been confirmed. Study I of this project investigated the anthropometric characteristics of cadaver femur anatomy.

The anterior cruciate ligament (ACL) may be considered the most important ligament of the knee (Markolf et al., 1995). The ACL prevents anterior translation of the tibia relative to the femur and also acts as a rotational stabilizer. ACL injuries are a common source of disability in the United States (Goris & Graf, 1996) and may begin a course of events to include instability, further ligamentous injury, meniscal injury, and arthritis (Markolf et al., 1995). Females are substantially more susceptible than males to suffer acute noncontact injury of the ACL (Arendt, 1994; Arendt & Dick, 1995; Hutchinson & Ireland, 1995; Loudin, Jenkins, & Loudin, 1996). Although the cause of disparity in injury rate is not fully understood, numerous extrinsic and intrinsic factors are believed to be relevant. Intercondylar notch geometry is one of the intrinsic factors suspected to predispose individuals to ACL injury and may differ between males and females. Anthropologists have established that men and women are morphologically different, but little is known about anthropometric differences regarding the intercondylar notch (Shelbourne, Davis, & Klootwyk, 1998).

Stenotic intercondylar notches have been implicated as a possible risk factor for ACL injuries. Two postulates prevail concerning intercondylar notch stenosis (Muneta, Takakuda, & Yamamoto, 1997). The first belief states that a narrow notch will have a correspondingly smaller ACL. A smaller ACL would be less strong and more likely to rupture. The second postulate states that a narrow notch surrounding a normal size ACL would provide insufficient space for the ligament to function normally. The reduced space could pinch the ACL and cause damage.

Notch width is most often reported as a Notch Width Index (NWI). The NWI is calculated as the ratio of intercondylar notch width to the width of the femoral condyles (Muneta, Takakuda, & Yamamoto, 1997; LaPrade & Burnett, 1994; Goris & Graf, 1996). NWI is further clarified in Figure 1-2. Values below 0.2 are considered stenotic and could predispose and individual to ACL injury. Data are inconclusive concerning differences in NWI between injured and uninjured individuals, as well as, males and females. Anderson and colleagues (1987) detected significant differences between ACL injured individuals and normal individuals when NWI was used as a basis of comparison. Souryal and Freeman (1993) deduced that athletes with stenotic intercondylar notches were more likely to suffer ACL injuries. LaPrade and Burnett (1994) studied 213 Division I collegiate athletes and concluded that intercondylar notch stenosis is associated with ACL injury, but no differences were in NWI were detected when compared by the sex of the athlete. Teitz, Lind, and Sacks (1997) compared data from individuals that had suffered noncontact ACL injuries with normal individuals and detected no differences between notch width indices for males and females or between normal healthy individuals and injured individuals. These authors also noted that a considerable amount of overlap exists for the ranges of NWI for males (0.19-0.32) and females (0.16-0.32).

Other researchers have used different approaches for analyzing the intercondylar notch. Houseworth and colleagues (1987) calculated the area of the intercondylar notch from roentgenograms of injured and uninjured individuals. The authors concluded that a narrow posterior arch of the intercondylar notch may be a risk factor related to ACL injury. Another unique study dealt with the cross-sectional area of the anterior cruciate ligament and the NWI. Muneta, Takakuda, and Yamamoto (1997) created molds of the ACL from 16 cadaver knees and concluded that the width of the intercondylar notch was not related to the size of the ACL and therefore, provides support for the second theory as a mechanism of ACL rupture. Another potentially critical aspect of the intercondylar notch is the shape of the notch itself. Intercondylar notch shapes may be classified as inverted U or A-shaped (Hutchinson & Ireland, 1995). Anderson et al. (1987) also observed the inverted U shape for normal notches, but stated that narrow notches tend to be more waveshaped and noted the need for additional research in the area of intercondylar notch shape.

Males and females incur ACL injuries at different rates (Arendt & Dick, 1995). No definitive evidence exists that explains this difference in injury frequency. Intercondylar notch geometry is considered a potential risk factor for ACL injury. Specifically, the NWI has been studied, but data are inconclusive. NWI may not be the only critical variable involved with intercondylar notch geometry because NWI does not provide details of notch configuration (Souryal & Freeman, 1993). Therefore, a more complete analysis of notch geometry should include configuration variables such as NAI and NSI. The purpose of this study was to determine if differences in intercondylar notch geometry exist between males and females. Specificially, notch width, notch area, and notch shape were compared between males and females.

Method

Participants

Two hundred human skeletons from the Terry Collection housed at the National Museum of Natural History, Smithsonian Institution in Washington, D.C., were used to establish the differences between males and females and to establish ranges of normal variation in intercondylar notch geometry. The Terry Collection was established in the early 20th century by Dr. Randall Terry at Washington University in St. Louis, MO, and consists of the skeletal remains of over 1,400 American individuals. Medical records (age at death, sex, race, living stature, living weight, and cause of death) are available for each individual. One hundred males (age at death 33.0 ± 10.3) and one hundred female skeletons (age at death 36.7 ± 11.8) of African and European descent were selected for inclusion in the study. Equal numbers of each race were included in each group. The individuals were between the ages of 18 and 55 years at the time of death and free from any orthopedic disorder as listed in medical records or observed upon initial inspection of the skeleton.

Materials

Digital photographs (640 x 480 pixels) were taken of the distal end of each femur. An Olympus D-200L digital camera was used for taking all photographs. The digital camera was fixed adjacent to a table and on a tripod (79 cm high and 22 cm from the table) and a fluorescent light was placed behind the camera in a manner that shed light on the condyles (see Figure 2-1). The camera was triggered remotely with a notebook computer. Images were downloaded to the notebook computer in JPEG format using Olympus Digital Camera software. Digital photographs were analyzed using MATLAB® with the Imaging Processing Toolbox. Data analysis was performed on a notebook computer using SPSS©. All equipment specifications appear in Appendix A.

Procedure

Digital photographs were taken of the distal end of the left femur for each test participant. The left femur was analyzed based on the assumption that the majority of the studied population was right-side dominant. If this assumption were true, the left femurs would have less variation due to the physical activity of the individual. Each femur was placed horizontally on the top surface of a table with the femoral condyles aligned with the edge of the table. Each photograph was analyzed using MATLAB for notch width index (NWI), notch area index (NAI), and notch shape index (NSI). The subroutine for making NWI, NAI, and NSI measurements is reproduced in Appendix B. NWI was

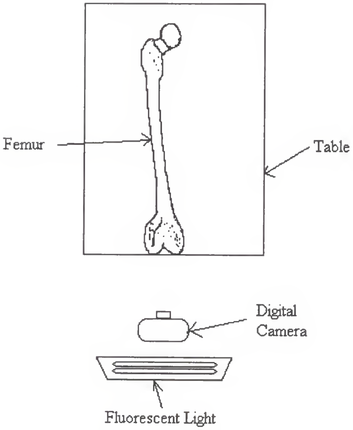
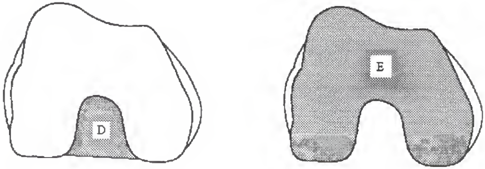


Figure 2-1. Overhead view of the data collection setup including digital camera, fluorescent lighting, and table.

calculated by dividing the width of the intercondylar notch by the width of the femoral condyles as shown in Figure 1-2. The widths of the intercondylar notch and femoral condyles were measured by drawing horizontal lines at approximately 2/3 of the anterior/posterior height of the intercondylar notch using a two-button PC mouse. NAI was computed by creating a ratio of the area encompassed by the intercondylar notch to the area of the femoral condyles. The area of the intercondylar notch was measured by using a series of mouse clicks (minimum of 10) to trace the outline of the notch. Condylar area was measured similarly (see Figure 2-2) by tracing the outline of the condyles with a minimum of 25 points. NSI was determined by dividing the width of the intercondylar notch by the height of the notch. The height of the intercondylar notch was measured by drawing a vertical line from the top of the notch to the most inferior level of the condyles as displayed in Figure 2-3. The same investigator completed all data collection during this study to eliminate any possibility of interrater reliability. Data from 20 participants were recollected to evaluate intrarater reliability.

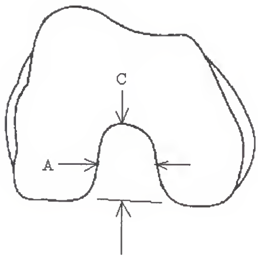
Design/Analysis

Descriptive statistics (means and standard deviations) were calculated for each of the three dependent variables (notch width index, notch area index, and notch shape index. Separate ANOVAs (Gender:2 x Race:2) were used to analyze the notch geometry data for each of the dependent variables. Intrarater reliability was assessed using the Pearson Product-Moment Correlation. All statistical tests were conducted with the conventional level of significance ($\alpha = .05$).



NAI = D/E

Figure 2-2. Notch Area Index (NAI) is calculated by dividing the area of the intercondylar notch (D) by the area of the condyles (E).



NSI = A/C

Figure 2-3. Notch Shape Index (NSI) is calculated by dividing the width of the intercondylar notch (A) by the height of the intercondylar notch (C).

Results

Data from 100 male participants and 100 female participants were analyzed. The three indices (NWI, NAI, and NSI) related to intercondylar notch geometry were calculated. Means (± SD) for each of the indices appear in Table 2-1 and Table 2-2. No difference was detected between males and females for NWI, $\underline{F}$ (1, 196) = 1.987, $\underline{p}$ = .160. NAI did not differ between males and females, $\underline{F}$ (1, 196) = 0.799, $\underline{p}$ = .372. NSI for males exceeded NSI for females, $\underline{F}$ (1, 196) = 8.835, $\underline{p}$ < .004. NWI for African descent individuals exceeded values for European descent individuals, $\underline{F}$ (1,196) = 5.217, $\underline{p}$ < .03. Skeletons of African descent had larger NAI than those of European descent, $\underline{F}$ (1,196) = 6.677, $\underline{p}$ < .02. NSI did not vary due to race, $\underline{F}$ = (1,196) = 3.668, $\underline{p}$ = .057. Pearson Product-Moment Correlation values all exceeded 0.90 and ranged from 0.932-0.949.

Discussion

The increased participation of females in organized sports since the inception of Title IX has coincided with an increase in orthopedic injuries. However, the increase in ACL injuries has not been proportional to the increase in participation. Females incur injuries to the ACL at much higher rates than their male counterparts (Loudin, Jenkins, & Loudin, 1996; Arendt & Dick, 1995). The difference in injury rates has fueled much research attempting to identify anatomical differences between males and females that may be related to function of the ACL. The results of previous research have been inconclusive.

The bony levers that compose the skeleton create the framework that dictates the movements of the musculoskeletal system. Therefore, malalignments of the skeletal

TABLE 2-1 INTERCONDYLAR NOTCH GEOMETRY INDICES BY GENDER

Index	Male	Female
NWI	0.255 ± 0.028	0.247 ± 0.032
NAI	0.173 ± 0.022	0.170 ± 0.025
NSI*	0.638 ± 0.089	0.599 ± 0.094

* = Significant difference ($p \leq .05$) between male and female.

TABLE 2-2 INTERCONDYLAR NOTCH GEOMETRY INDICES BY RACE

Index	African Descent	European Descent
NWI*	0.257 ± 0.028	0.247 ± 0.032
NAI*	0.176 ± 0.024	0.167 ± 0.023
NSI	0.606 ± 0.087	0.631 ± 0.098

* = Significant difference ($p \leq .05$) between African descent and European descent.

system have been implicated as possible risk factors for ACL injuries. More specifically, the geometry of the intercondylar notch of the femur may lead to an increased incidence of ACL tears and could differ between males and females. Data presented by Shelbourne, Davis, and Klootwyk (1998) support this belief and suggest that ACL injury rates are affected by intercondylar notch width and that females have narrower notches than males when grouped by the height of the individual. Souryal and Freeman (1993) reported similar findings using the more conventional measure of NWI. Both studies analyzed the size of the intercondylar notch using radiographs. Another study of NWI indicated that males and females do not differ statistically, although females tended to have smaller notches than males (Teitz, Lind, & Sacks, 1997). Additionally, LaPrade and Burnett (1994) used radiographs to study 213 Division I collegiate athletes and concluded that intercondylar notch stenosis is associated with ACL injury, but no differences in NWI were detected when compared by the sex of the athlete.

The width of the intercondylar notch may not be the only critical variable related to ACL injury. Another approach has been used to analyze the notch. Houseworth and colleagues (1987) used radiographs to calculate the area of the intercondylar notch and concluded that a posterior arch of the intercondylar notch with small area may be a risk factor related to ACL injury. No other studies have analyzed the area encompassed by the intercondylar notch.

The shape of the intercondylar notch may also play a role in ACL injuries. Observational research has indicated that narrow and wide notches have different shapes. Narrow notches tend to be more A-shaped or wave-shaped while wider notches tend to be more round or shaped like an inverted U (Tanzer & Lenczner, 1990; Anderson et al.

1987). No studies have attempted to quantify the shape of the intercondylar notch. The purpose of this study was to identify quantitative differences in intercondylar notch geometry between males and females without history of knee injury. In the present work, a common measure of notch width (NWI) and 2 novel measures of notch geometry (NAI and NSI) were calculated for comparison.

NWI did not vary between male and female cadaver femora. The average values for males and females (0.26 and 0.25, respectively) greatly exceeded the value of 0.2 which is considered stenotic. However, the entire sample of males and females included some values below 0.2. Souryal and Freeman (1993) calculated similar values for NWI (males = 0.239 ± 0.040, females = 0.217 ± 0.041). Comparable measures were made by Muneta, Takakuda, and Yamamoto (1997) for males (0.25) and females (0.28). The data collected here indicate that intercondylar notch width does not vary between males and females and therefore, can not explain the discrepancy in ACL injury rates. The results agree with the work of Teitz, Lind, & Sacks (1997). The increased rate of ACL ruptures in females can not be solely attributed to NWI.

NAI did not vary between males and females. Thus, the anterior cruciate ligament should pass through the same relative two-dimensional space regardless of the sex of the individual. Thus, similar to NWI, NAI can not be responsible for the difference in ACL injury rates between males and females.

NSI was greater for males than females. NSI is a relative measure of notch width in the medial/lateral direction to the notch height in the anterior/posterior direction. NWI did not vary between males and females, but apparently the width of the notch normalized by the height of the notch is greater in males. This indicates that the shape of the notch in males may be more round than in females. Knees with lower NSI could be detrimental to normal function of the ACL. When the knee is in full extension, the cruciate ligaments are pulled tight and will reside in the more anterior portion of the intercondylar notch. A low NSI indicates that this particular region of the intercondylar notch will be more narrow and thus will provide less space for the ligament to function correctly. In fact, Norwood and Cross (1977) have noted that the ACL is in direct contact with the intercondylar shelf when the knee is in full extension. The difference in shape of the intercondylar notch may partially explain the difference in injury rates between males and females. Normally, the dimensions of the intercondylar notch are such that little space is not filled by the cruciate ligaments (Tanzer & Lenczner, 1990). Tanzer & Lenczner (1990) also noted that narrow notches may be more wave-shaped while wider notches tend to be more round. A notch that is relatively wider and more round may provide more space for the ACL when the knee is near or in full extension.

As a final point, the differences in intercondylar notch geometry between the two races are perplexing. Epidemiological studies of knee injuries have not differentiated between the races relative to frequency of occurrence. Thus, no race hypotheses were proposed for the anatomical comparisons. More importantly and corresponding to the primary research question, hypotheses were generated for gender differences. NWI and NAI for individuals of African descent were greater than the same indices calculated on the European descent individuals while NSI did not vary between races. These results indicate that people of European descent could be more predisposed to ACL injury than their African descent counterparts because less room is available in the intercondylar notch for normal movement of the ACL. However, there are no empirical data to indicate that any differences exist in ACL injury rate between individuals of African and European descent.

Conclusions

Traditional measures of intercondylar notch geometry do not vary between males and females without history of ACL injury. However, a new measure of notch shape (NSI) does vary between males and females. These data may help to explain the higher rate of ACL injury observed in females. NSI, as calculated here, is an initial attempt to quantify the shape of the intercondylar notch. The results indicate a need for more exact quantification of the shape of the notch. The difference in ACL injury rates between males and females is most likely a multifactorial problem. The data collected in this study may be used in conjunction with other measures of musculoskeletal alignment and knee kinematics to obtain a more complete analysis of the issue.

CHAPTER 3
LIVING PARTICIPANT ANTHROPOMETRICS: DO MALES AND FEMALES DIFFER ON MUSCULOSKELETAL ALIGNMENT?

Study II supplements data collected in Study I. Initially, cadaveric skeletal geometry of the intercondylar notch were analyzed to discern differences between males and females that could be related to ACL injury rates. This study examined musculoskeletal alignment of living individuals in an attempt to detect differences between males and females that may be related to ACL injury.

Static musculoskeletal alignment has been implicated as predisposing individuals to injury of the ACL. The ACL prevents anterior translation of the tibia relative to the femur and acts as a rotational stabilizer. ACL injuries are a common source of disability in the United States (Goris & Graf, 1996) and may begin a course of events that include instability, further ligamentous injury, meniscal injury, and arthritis (Markolf et al., 1995).

Females are much more likely to suffer acute noncontact injury of the ACL than males (Arendt, 1994; Arendt & Dick, 1995; Hutchinson & Ireland, 1995; Loudin, Jenkins, & Loudin, 1996). The cause(s) of the difference in injury rate is not fully understood. However, numerous extrinsic and intrinsic factors are believed to be associated with ACL injury. The focus of this study involves intrinsic factors, specifically Q-angle, thigh foot angle, and subtalar joint ratio, which may differ between males and females and could be related to ACL injury.

Q-angle

The Q-angle is defined as the acute angle formed by a line drawn from the center of the tibial tuberosity through the midpoint of the patella and a line running from the midpoint of the patella to the anterior superior iliac spine (Schulthies, Francis, Fisher, & Van De Graaff, 1995). As shown in Figure 1-3, the Q-angle is measured with the knee in full extension and is intended to provide an estimate of the direction of the resultant force produced by the quadriceps (Schulthies et al., 1995; Terry, 1980). Large Q-angles have been associated with lower limb injuries and knee extensor dysfunction (Cowan et al., 1996; Ciullo, 1993; Hunter, 1984). Theoretically, a greater Q-angle increases the horizontal component of the quadriceps muscle force and pulls the patella laterally. Increased horizontal force on the patella exposes the ligaments and menisci of the knee to greater loads and renders these structures vulnerable to injury (Hsu et al., 1990).

The majority of the Q-angle research has indicated that females have larger Q-angles than their male counterparts (Hsu et al., 1990; Woodenlang L. H. & Francis, R. S., 1992). Reports indicate that the Q-angle for females ranges from 2.0° to 8.5° greater than males (Agletti, Insall, & Cerulli, 1983; Hvid, Andersen, & Schmidt, 1981). Conversely, Livingston and Mandigo (1997) detected no significant differences in Q-angle between males (10.5°) and females (12.2°). However, no research has been reported in which the mean Q-angle in males exceeded that for females (Livingston, 1998). To date, studies concerning Q-angle differences between males and females are inconclusive.

Thigh Foot Angle

Thigh foot angle (TFA) has recently been linked with injuries to the lower limb and specifically ACL injuries (Heugel et al., 1999). TFA is a combined measure of tibiofibular torsion and alignment of the structures distal to the talocrural joint (Stuberg, Temme, Kaplan, Clarke, & Fuchs, 1991). TFA is measured by positioning the individual in a prone position, flexing the knee to 90°, moving the ankle to a neutral position, and determining the angle formed by a line drawn from the midpoint of the calcaneus and the second toe and the long axis of the femur (Cheng, Chan, Chiang, & Hui, 1991).

Heugel et al. (1999) compared TFA values between ACL-injured and uninjured female athletes and observed that individuals who had suffered ACL injuries had significantly higher TFA values (23.3°) than those that did not have ACL injuries (20.0°). Excessive internal tibiofemoral rotation may also result from excessive external tibial torsion, measured as TFA. Heugel et al. (1999) suggest that an increase in TFA may cause poor foot plant in a position of increased external rotation relative to the knee. Tiberio (1987) noted that external tibial torsion should be considered an indirect factor that leads to excessive pronation of the subtalar joint. A possibility exists that, as the foot lands in external rotation, the forward momentum of the leg may force the subtalar joint into pronation as the center of mass of the entire body moves anteriorly over the foot.

Subtalar Joint Axis

The bony geometry of the foot and ankle has also been implicated as a risk factor for injuries to the lower limb. Specifically, excessive foot eversion and abnormal tibial rotation have been associated with knee injuries (Hintermann & Nigg, 1993). The relationship between these two movements is determined by the subtalar joint axis. The subtalar joint is defined as the articulation between the talus superiorly and the calcaneus and navicular inferiorly (Rockar, 1995). The subtalar joint axis passes through the talus and extends from below on the lateral side of the heel, upward, forward, and medially

(Inman, 1976; Hicks, 1953; Manter, 1941). The mean axis lies 42° above the horizontal plane and 16° medial to the midline of the foot, as shown in Figure 1-4 (Manter, 1941). However, great variation in the location of the axis exists among different individuals. A lower axis will result in greater frontal plane movement, which will occur as calcaneal eversion and inversion. A higher axis will result in greater horizontal plane movement, which occurs as internal and external tibial rotation. Therefore, a lower axis accompanied by greater eversion of the foot may predispose an individual to greater injuries of the foot, while a higher axis with greater tibial rotation may predispose an individual to more knee injuries (McClay & Bray, 1996).

The relationship between tibial rotation, which is a transverse plane motion, and inversion/eversion of the subtalar joint, which is a frontal plane motion, is not fixed (Root, Orien, & Weed, 1977). Depending on the height of the subtalar joint axis, some individuals may have proportionally more internal tibial rotation with a fixed amount of eversion than others. Subtalar joint pronation is a triplanar motion with movement occurring around a single axis, which is oblique to the three cardinal planes. As with any planar motion, the movement occurs in the plane that is perpendicular to the joint axis. Normally, the subtalar joint axis lies closest to the sagittal plane, thus, the majority of motion occurs in the frontal and transverse planes. A high subtalar joint axis lies closer to the frontal plane and limits movement in that plane; therefore, all compensatory motion must occur in the transverse plane as rotation. Individuals with high subtalar joint axes may be more at risk of ACL injury upon compensatory pronation of the subtalar joint than those with lower subtalar joint axes because they have a greater amount of internal tibial rotation with a given amount of subtalar joint pronation (see Figure 1-5).

Several structural elements have been considered as risk factors for ACL injury. However, no singular musculoskeletal factor has appeared as the primary risk factor. This study was intended to complement the skeletal data collected in Study I with musculoskeletal data. These data may resolve inconsistencies in previous research regarding gender differences and musculoskeletal alignment. Currently, there are no accurate methods that have been developed to measure the location of the subtalar joint axis nor are there data comparing the location of the subtalar joint axis between males and females. The goals of this project are as follows: (a) evaluate subtalar joint function with a manufactured clinical device, (b) compare Q-angle, (c) TFA, and (d)subtalar joint function between males and females.

Method

Participants

Fifty-seven male and female volunteers were recruited from graduate and undergraduate classes at the University of Florida and the surrounding community to participate in this study. Individuals were required to be physically active (participating in physical activity at least two times a week) to qualify. Participants had no history of lower extremity injury in the 6 months prior to volunteering for testing.

Materials

Each participant's weekly physical activity level was determined by questionnaire (Appendix C). Q-angle measurements and TFA measurements were performed with a modified goniometer. The goniometer was fitted with a clear plastic extension (2.9 cm wide, 75.6 cm long, 0.32 cm thick) on one arm. A 10.2 cm diameter angle finder was used to measure calcaneal range of motion. A laser pointer was used in conjunction with a tripod to measure tibial rotation. Participants stood on a 40 cm wooden platform while subtalar joint function was tested. Appendix A contains specifications for all equipment used in this portion of the study.

Procedure

Before testing began, each participant read and signed an informed consent agreement, as required by the Institutional Review Board of the University of Florida (see Appendices D and E). Participants removed their shoes and socks and were fitted with black tight fitting shorts. At this point, a series of anthropometric measurements were taken. The same investigator made all anthropometric measurements.

The Q-angle was measured with the participant standing upright with the knees fully extended. The participant's feet were positioned at shoulder width and parallel to each other. The goniometer was placed with the axis of rotation over the center of the patella. The short arm of the goniometer was then aligned with the tibial tuberosity and the long arm of the goniometer was aligned with the anterior superior iliac spine and the angle was recorded. This procedure was repeated three times with each valuerecorded and then averaged.

TFA was measured by positioning the individual in a prone position, flexing the knee to 90°, and moving the ankle to a neutral position by pressing on the head of the 5th metatarsal head until the bottom of the foot was horizontal. The axis of the goniometer was then placed at the center of the calcaneus with the short arm of the goniometer aligned with the 2nd toe and the long arm of the goniometer aligned with the long axis of the femur. The technique is displayed in Figure 3-1. The angle was then read from the

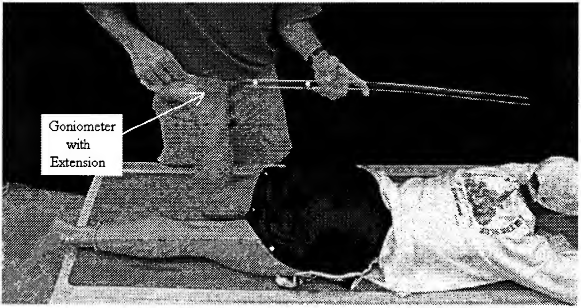


Figure 3-1. Experimental setup for measuring thigh foot angle. Participants were positioned prone with their knee flexed to 90°. The angle between the long axis of the femur and a line from the center of the calcaneus to the second toe were measured using a modified goniometer.

goniometer. Again, this procedure was completed three times and the resulting values were recorded and averaged.

Subtalar joint function was evaluated by determining the subtalar joint movement ratio. The subtalar joint movement ratio was calculated by dividing the amount of tibial rotation by the amount of calcaneal inversion and eversion. Tibial rotation and calcaneal movements were measured simultaneously. During data collection the participant stood on a 40 cm high wooden platform without shoes. The calcaneal range of motion was measured using an angle finder fixed to the posterior aspect of the individual's calcaneus (in the frontal plane). A second angle finder was fixed to the posterior part of the lower leg to measure the angle of the lower leg relative to absolute vertical. Tibial rotation was measured by recording the horizontal displacement of a laser beam directed onto a flat surface. A laser pointer was fixed to the medial portion of the test participant's tibia using elastic straps fitted with Velcro™ fasteners. A tripod with a grid for determining angular displacement of the tibia was then oriented perpendicular to the beam produced by the laser pointer at a distance of 1 m from the tip of the pointer as shown in Figure 3-2. At the 1 m distance, a horizontal displacement of 1.75 cm on the grid was equivalent to 1° of tibial rotation. The participant then everted their calcaneus maximally followed by a maximal inversion with the forefoot remaining in contact with the platform. While the participant held their calcaneus in maximal eversion, the experimenter recorded the angular displacement displayed on the heel-mounted angle finder and the angular displacement displayed on the tibial rotation grid. The ratio representing the function of the subtalar joint was calculated as follows:

$$r = [IR - ER]/[EV - INV].$$

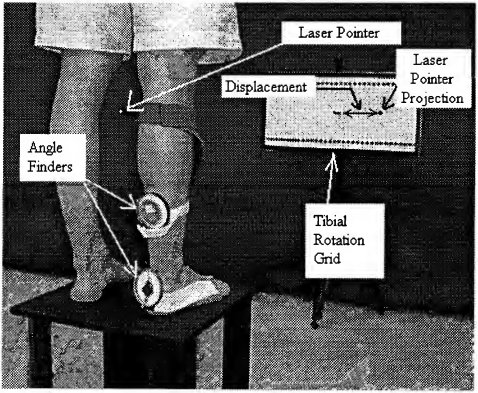


Figure 3-2. Experimental setup for measuring subtalar joint movement. Participants stood on a 40 cm high wooden platform while wearing two angle finders and a laser pointer. During maximal pronation and supination, rotational values were read from the tibial rotation grid and the angle finders were recorded.

Where EV represents the maximum calcaneal eversion relative to the tibia, INV represents the maximum calcaneal inversion relative to the tibia, IR represents the maximum internal rotation of the tibia, and ER represents the maximum external rotation of the tibia. The process was conducted three times and averaged. The same investigator made all measures of Q-angle, TFA, and subtalar joint ratio. Data were analyzed on a notebook computer using the SPSS® statistical software.

Design/Analysis

Means and standard deviations were calculated for each of the dependent variables (Q-angle, thigh foot angle, and subtalar joint ratio). Independent t-tests were used to measure differences between males and females across the dependent variables. The original level of significance for all tests was set at .05. The likelihood of making a Type I error was reduced by using the Bonferroni adjustment for multiple comparisons, resulting in an adjusted level of significance ($\alpha = .017$).

Results

Participant Characteristics

Fifty-seven individuals participated in the study. The self assessment of physical activity confirmed that the participants were physically active two to three times per week. A summary of their characteristics appears in Table 3-1.

Anthropometrics

Measures of musculoskeletal alignment were made on each of the participants. Q-angle, TFA, and subtalar joint ratio values are displayed in Table 3-2. Q-angle in females exceeded values recorded for their male counterparts, $\underline{t}$ $(55) = 4.82$, $\underline{p} < .001$. However,

TABLE 3-1 CHARACTERISTIC INFORMATION FOR TEST PARTICIPANTS

Characteristic	Male	Female
n	30	27
Age (yr ± SD)	21.3 ± 2.4	20.6 ± 1.0
Mass (kg ± SD)	78.1 ± 12.5	58.9 ± 9.2
Height (cm ± SD)	176.4 ± 6.2	162.7 ± 6.0

TFA and subtalar joint ratio did not vary between males and females, $\underline{t}$ (55) = 2.40, $\underline{p}$ = .020; $\underline{t}$ (55) = 0.57, $\underline{p}$ = .573.

Discussion

The inception of Title IX has led to increased participation of females in organized sports and a corresponding increase in orthopedic injuries. However, the increase in ACL injuries has not been commensurate to the increase in participation. Females incur injuries to the ACL at much higher rates than their male counterparts (Loudin, Jenkins, & Loudin, 1996; Arendt & Dick, 1995). The discrepancy in ACL injury rate is a significant problem and has provided the impetus for research attempting to identify anatomical differences between males and females that may be related to function of the ACL. Unfortunately, previous findings have been inconclusive. The bones, muscles, and ligamentus structures that compose the musculoskeletal system dictate the mechanics of locomotion. Thus, malalignments of the musculoskeletal system have been implicated as possible risk factors for ACL injuries.

Q-angle

The Q-angle is an estimate of the direction of pull of the quadriceps muscle (Schulthies et al., 1995). Abnormal Q-angle values have been implicated in several disorders of the knee (Horton & Hall, 1989). High values of Q-angle (> 15-20°) are thought to contribute to knee extensor dysfunction (Livingston, 1998) and may be related to ACL injury. A larger Q-angle may increase the horizontal component of the quadriceps muscle force and pull the patella laterally. The increase in horizontal force on the patella may endanger the ligaments and menisci of the knee (Hsu et al., 1990).

TABLE 3-2 MEASURES OF MUSCULOSKELETAL ALIGNMENT

Alignment Variable	Male	Female
Q-angle (° ± SD)*	13.1 ± 3.0	17.5 ± 3.8
TFA (° ± SD)	8.8 ± 5.2	12.4 ± 5.9
Subtalar joint ratio ± SD	2.35 ± 1.7	2.10 ± 1.6

* = Significant difference ($p \leq .017$) between male and female.

The Q-angle values measured in this study are similar to previously reported data. Horton and Hall (1989) studied 100 individuals with no history of knee disorders and computed mean Q-angle values for females (15.8 ± 4.5°) and males (11.2 ± 3.0°). Hvid and Anderson (1982) examined people with patellofemoral pathologies and concluded that females had larger Q-angles (20°) than their male counterparts (12°). Aglietti, Insall, and Cerulli (1983) also noted that healthy females had larger Q-angles (17 ± 3°) than males (14 ± 3°). The current study contributes to the body of knowledge that females have larger Q-angles than males, but there are no data to indicate why the difference exists (Horton & Hall, 1989).

The importance of the difference in Q-angle between genders may surface relative to a mechanism of ACL injury. ACL injuries most often occur in a noncontact fashion resulting from a combination of flexion and internal rotation of the tibia (Hess et al., 1994; Jarvinen et al., 1994). Feagin, Cabaud, and Curl (1982) also noted the importance of transverse kinematics of the knee and observed that the ACL actually stretches over the lateral femoral condyle as the tibia rotates internally relative to the femur. Others have hypothesized that a relationship may exist between Q-angle and rearfoot motion (Kernozek & Greer, 1993). Specifically, a larger Q-angle may result in greater subtalar joint pronation (Subotnick, 1975). Greater subtalar joint pronation may cause excessive internal rotation of the tibia and increase the distance between the attachment points of the ACL producing tension in the ligament (Ahmed, Burke, Duncan, & Chan, 1992). Therefore, excessive internal rotation of the tibiofemoral joint partially resulting from a large Q-angle may predispose ACLs to injury. Larger Q-angle values observed in the females studied may help to explain the difference in ACL injury rates between the genders.

Thigh Foot Angle

Thigh foot angle is a combined measure of tibiofibular torsion and the alignment of the foot (Stuberg et al., 1991). The bony alignment of the lower extremity, including high values of TFA, has been implicated as predisposing to overuse injuries (Krivickas, 1997) and ACL rupture (Heugel, 1998). Excessive TFA may lead to a poor foot plant position as hypothesized by Heugel (1998) or could cause a disproportionate amount of pronation and commensurate internal rotation of the tibia and possible internal tibiofemoral rotation.

The current TFA data conflict with earlier TFA studies. Stuberg and colleagues (1991) measured TFA goniometrically on 17 normal individuals and compared the values collected by three different investigators ($15.1 \pm 4.3°$, $14.3 \pm 5.0°$, and $17.6 \pm 5.1°$). The same goniometric techniques were used in both studies, but the participant populations were quite different. Stuberg's participants ranged in age from three to 24 and were only 17 in number. However, the present study examined 57 individuals who varied in age from 19 to 31 years. Heugel et al. (1999) examined individuals in the same range as the current work, but reported higher values of TFA for ACL injured (23°) and uninjured (20°) Division I athletes. The discrepancy may be attributed to the differences between elite athletes and healthy physically active individuals. A third study considered the lower limb alignment of over 2000 Chinese children. Cheng and colleagues (1991) observed that two-year-old children had a mean TFA of 15° that increased to a value of 35° by age

12. However, the authors did note that Chinese children tend to have TFA measures 10-15° greater than children of other races of similar age.

Excessive internal tibiofemoral rotation may result from excessive external tibial torsion, measured as TFA. No significant differences were detected between normal males and females concerning TFA although there is strong trend suggesting that females may have larger values of TFA than their male counterparts. The 3.6° difference in TFA mean values between males and females may not be statistically significant, but could be biomechanically significant. The "biomechanical" difference actually represents an increase in TFA of 40% for the females tested relative to their male counterparts. TFA is most likely not solely responsible for the difference in ACL injury rates between males and females, but could be a contributing factor. Perhaps, an excessive TFA combined with other factors may predispose an individual to a mechanism of ACL injury.

<u>Subtalar Joint Ratio</u>

Movements that may lead to ACL injury often begin with footstrike. The foot and ankle are the first part of the kinetic chain that contacts the ground and effect the movements that occur further up the kinetic chain at the knee. Specifically, the amount of tibial rotation that occurs is determined by the location of the subtalar joint axis. If this axis lies closer to the frontal plane, an increased amount of internal tibial rotation will occur for any given amount of calcaneal eversion. Therefore, an individual with a high subtalar joint axis may be predisposed to internal tibiofemoral rotation and possibly ACL injury during periods of excessive pronation.

Unfortunately, no simple, noninvasive, clinical methods have been developed to determine the location of the subtalar joint axis. The location of the subtalar joint axis has been determined experimentally using cadavers to be 42° above the horizontal plane and 16° medial to the midline of the foot (Inman, 1976; Hicks, 1953; Manter, 1941). McClay and Bray (1996) used radiographs to characterize foot structure in an attempt to identify a method for measuring the location of the subtalar joint axis and noted that two of their measures were similar to previously reported data, but may not correlate with the actual orientation of the axis. Other researchers attempted to locate the subtalar joint axis clinically by creating equations based on non-weight bearing measurements (van den Bogert, Smith, & Nigg, 1994; Phillips & Lidtke, 1992). The movements of the subtalar joint are most important during closed-chain activities, therefore, the results of non-weight bearing tests may not be relevant to the study of dynamic activities which may lead to injury.

An alternative to actually locating the subtalar joint axis may be to determine the net effect of the axis. A ratio of tibial rotation to calcaneal movement could be a better descriptor of subtalar joint function. If the axis lies nearly halfway between the transverse and frontal planes, then the amount of tibial rotation should equal the amount of calcaneal inversion/eversion. Thus, a ratio of these movements would equal zero. The results reported here do not support this theory. The computed subtalar joint ratios do not vary between genders (2.35 ± 1.7 for males vs. 2.10 ± 1.6 for females), but greatly exceed the theoretical value of 1 that would result from previous in-vitro and in-vivo studies. This indicates that individuals have much greater tibial rotation for any amount of calcaneal movement than previously hypothesized. This finding may be particularly important relative to the prior discussion of a common mechanism of ACL injury (internal tibiofemoral rotation). A high subtalar joint ratio may result in a greater amount of internal tibial rotation for any amount of calcaneal eversion. This situation could stress the ACL as a result of internal tibiofemoral rotation and create a situation in which structural damage is a possibility. Based on epidemiological data and the increased rate of ACL injury for females, it was hypothesized that females would have a higher subtalar joint ratio than males. However, males and females did not vary according to subtalar joint ratio. Therefore, the subtalar joint ratio can not help to explain the difference in ACL injury rates between genders, but a high ratio may be a measurable risk factor for any individual regardless of gender.

Conclusions

Several malalignments of the musculoskeletal system have been implicated as risk factors that predispose individuals to ACL injury and may differ between males and females. Abnormally high values of Q-angle, thigh foot angle, and subtalar joint movement ratio may be predisposing factors. However, the only statistical difference identified in the current study was the Q-angle. The possible relationship between Q-angle and rearfoot motion may help to explain the difference in ACL injury rates between genders. Subotnick (1975) stated that a larger Q-angle may lead to greater subtalar joint pronation. As reported here, females have Q-angles 4.4° larger than males and therefore, may be predisposed to more subtalar joint pronation and simultaneous internal tibial rotation. Although no statistical differences were detected between the groups for subtalar joint ratio, the new measure of subtalar joint ratio may be a better technique of understanding subtalar joint function than traditional measures for locating the subtalar joint axis. Possible links between these static measures of musculoskeletal alignment and the kinematics of knee movements were explored in Study III.

CHAPTER 4
A COMPARISON OF KNEE KINEMATICS BETWEEN MALES AND FEMALES

The third study of this project was designed as an extension of Studies I and II. In the first two studies, anatomical characteristics believed to be related to ACL injuries were compared between males and females. Study III was an examination of a common mechanism of ACL injury (tibiofemoral rotation). The amount of tibiofemoral rotation during running and landing was compared between males and females. These data were combined with the anthropometric data from Study II in an attempt to identify any possible relationships between anthropometrics and this mechanism of injury.

Certain bony configurations of the foot and ankle that result in atypical kinematics have been implicated as risk factors for injuries to the lower limb. Specifically, excessive foot eversion and abnormal tibial rotation have been associated with knee injuries (Hintermann & Nigg, 1993). The relationship between these two movements is determined by the subtalar joint axis. Great variation in the location of the axis exists among different individuals. A lower axis will result in greater frontal plane movement, which will occur as calcaneal eversion and inversion. A higher axis will result in greater horizontal plane movement that will occur as internal and external tibial rotation. Therefore, a lower axis accompanied by greater eversion of the foot may predispose an individual to greater injuries of the foot, while a higher axis resulting in greater tibial rotation may predispose an individual to more knee injuries (McClay & Bray, 1996). Specifically, excessive internal rotation of the tibia increases the distance between the attachment points of the ACL producing tension in the ligament (Ahmed, Burke, Duncan, & Chan, 1992).

Increased tension resulting from excessive tibiofemoral rotation may be detrimental to the ACL, which is considered the most important ligament of the knee (Markolf et al., 1995). The ACL prevents anterior translation of the tibia relative to the femur and also acts as a rotational stabilizer. ACL injuries are a common source of disability in the United States (Goris & Graf, 1996), and may begin a course of events to include instability, further ligamentous injury, meniscal injury, and arthritis (Markolf et al., 1995). Females are substantially more susceptible than males to suffer acute noncontact injury of the ACL (Arendt, 1994; Arendt & Dick, 1995; Hutchinson & Ireland, 1995; Loudin, Jenkins, & Loudin, 1996).

Mechanisms of Injury

Sports involving physical contact among players are expected to produce ACL injuries (Huston & Wojtys, 1996). However, the majority of ACL injuries occur in a noncontact fashion. Noyes and colleagues (1983) reported that 78% of ACL injuries were noncontact and occurred during landing from a jump, cutting, or sudden deceleration. In a survey of ACL injured skiers, 88% of the injuries occurred in the presence of tibiofemoral rotation (Jarvinen et al., 1994). The two primary mechanisms were a combination of valgus knee position and external rotation and a combination of flexion and internal rotation. The predominant injury mechanism for the female skiers studied was flexion and internal rotation (Jarvinen et al., 1994). Hess and colleagues (1994) studied 151 sports-related ACL injuries and observed a similar trend with the majority of injuries occurring as a result of flexion and internal tibial rotation. Emerson (1993) noted the prevalence of ACL injuries in basketball players as the result of deceleration and change of direction when the tibia is internally or externally rotated. Tibial rotation appears to be involved in the majority of non-contact ACL injuries.

The close relationship between tibial rotation, subtalar joint pronation, and malalignments of the foot suggests that management of the compensatory motions through foot orthotic intervention may control excessive motions and reduce the risk of knee injury. Foot orthotics are commonly used to control subtalar joint movement and are intended to restore dynamic stability and reduce compensatory pronation of the subtalar joint during the initial stance phase of gait (McCulloch, Brunt, & Vander Linden, 1993). The purpose of this work was to determine the amount of tibiofemoral rotation produced during dynamic activities and to detect any differences between males and females as they wore different in-shoe orthotics. In addition, a prediction equation for tibiofemoral rotation based on three anthropometric factors, gender, and task was computed. Finally, the influence of orthotics on tibiofemoral rotation was measured. Hypotheses included the following: (a) females would have greater internal tibiofemoral rotation than males, (b) orthotics would influence tibiofemoral rotation, and (c) tibiofemoral rotation could be predicted based on gender, physical task, and anthropometric measures.

Method

Participants

Fifty-one males and females volunteered to participate. They were recruited from graduate and undergraduate classes at the University of Florida. Individuals were required to be physically active by participating in strenuous activity at least two times a week. All participants were free from lower extremity injury with no injury in the previous 6 months that prevented participation in training or competition.

Materials

A physical activity questionnaire was administered to all participants to verify the level of physical activity maintained by each of the participants. Q-angle measurements and TFA measurements were performed with a modified goniometer. A large goniometer was fitted with a clear plastic extension (2.9 cm wide, 75.6 cm long, 0.32 cm thick) on one side. An altered 10.2 cm diameter angle finder was used to measure calcaneal range of motion. A laser pointer was used in conjunction with a tripod to measure tibial rotation (see Figure 3-1). Participants stood on a 40 cm high wooden platform during the subtalar joint function portion of testing. Running trials were completed on a treadmill. Two pointing devices, modified from the version described by Cornwall and McPoil (1995), were used in the study. Each pointer consisted of a 37 mm diameter wooden sphere with a concave section of the sphere removed and a 64 mm square piece of thin steel fixed to the concave section of the sphere. The steel was intended to provide a larger surface area of contact between the pointing device and the leg of the participant. Two 56 mm long aluminum rods were fixed to the sphere and oriented at 90° to each other. A 22 mm diameter reflective sphere was attached to the end of each rod. The orthogonal pointing device is depicted in Figure 4-1. During testing, one pointing device was attached to the distal femur and the second to the proximal tibia using Velcro™ straps. High-speed video data (500 Hz) were collected using a Kodak Motion Corder Analyzer. A set of 500 W halogen lamps provided light for high-speed video collection. Standard video data

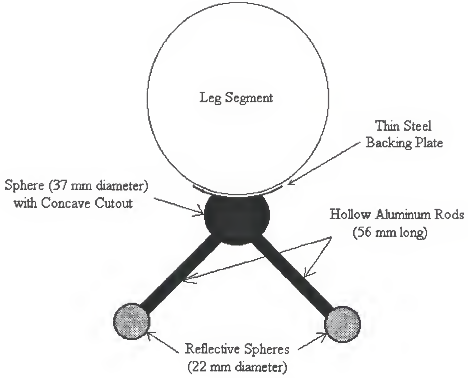


Figure 4-1. Top view of one of two orthogonal pointing devices that were attached to the anterior portions of the distal end of the femur and the proximal end of the tibia using Velcro™ straps.

(30 Hz) were collected using a JVC video camera. High-speed video data were analyzed using the automatic marker identification routine on the PEAK5 motion analysis system. Coordinate data were then downloaded in ASCII format for further analysis. During dynamic testing, participants wore identical athletic shoes and in-shoe orthotics. One pair of each size of orthotic was unaltered and two pairs of each size were modified by adding 8° high-density polyethylene posting material to inferior surface of the orthotic on either the medial or lateral side. The locations and types of posting material appear in Figure 4-2. Equipment specifications appear in Appendix A.

A Note on Marker Movement. The goal of movement analysis is to reconstruct the instantaneous spatial relationships of bony anatomical landmarks, regardless of the ability of the experimenter to actually observe the points during the movement (Cappozzo et al., 1997). The most direct way to measure skeletal motion is to insert pins directly into the bones. However, direct measures of skeletal motion using markers attached to pins are invasive and not suitable for routine analyses (Reinschmidt et al. 1997). Routine video analyses are normally performed by applying reflective markers on the skin over anatomical landmarks of interest and capturing video images while a dynamic task is performed. However, markers on the skin are not stationary relative to the underlying bony landmarks (Cappozzo et al., 1997). Ideally, a correction factor could be created to account for the motion of the skin relative to the bony structures of interest. Unfortunately, correction algorithms have not been realized because marker movement errors are highly participant dependent (Reinschmidt et al., 1997). Therefore, video data from this project were collected and analyzed without the use of a correction algorithm. It is understood that the absolute magnitudes of movement may not be correct for video

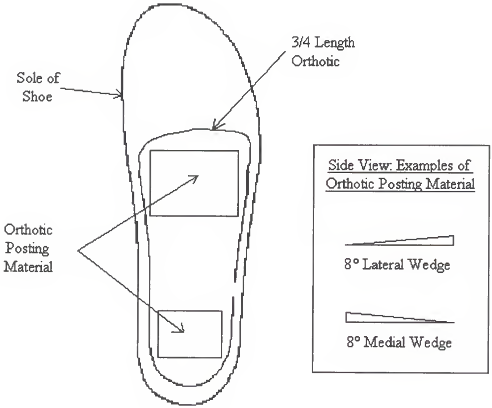


Figure 4-2. Orthotics were posted in the heel and forefoot regions with either 8° medial posting material or 8° lateral posting material.

analysis based on skin markers, but relative changes in movement for different test conditions are important and measurable.

Procedure

Before testing began, each participant was required to read and sign an informed consent agreement, which was approved by the Institutional Review Board of the University of Florida. Participants were then fitted with black tight-fitting shorts and asked to remove their shoes and socks. At this point, a series of anthropometric measures were made. The same investigator made all anthropometric measurements. Q-angle, TFA, and subtalar joint movement data collected in Study II were used in this experiment to create a prediction equation for tibiofemoral rotation.

Tibial and femoral orthogonal pointing devices were securely fastened to their respective limb segments. Each pointing device was attached with the rods parallel to the ground during standing and on the anterior portion of the limb segment. At this point the participants were fitted with identical athletic shoes. Standardized athletic shoes were intended to control any differences that could have arisen from differences in shoe design or excessive wear. The four levels of orthotic intervention (no orthotic, standard orthotic, medially posted orthotic, and laterally posted orthotic) were administered using a Latin-Square design. The next series of activities were performed with each of the four orthotic interventions. Each participant was initially assigned to one of the two activities (running or landing on one leg after stepping off a small platform). The order of the activity was counterbalanced. The running task consisted of a 5-minute warm-up on the treadmill followed by four 3-minute trials. Each trial was performed with a different orthotic-shoe combination. The treadmill remained level at all times. During the warm-up period, the individual selected a comfortable velocity comparable to their normal running pace. This velocity was recorded and used during subsequent running trials under the orthotic conditions. Four 3-minute trials were performed at the warm-up speed. Each trial was conducted at a different level of orthotic intervention and a short break was given between trials for the changing of orthotics. During the running activity, high-speed video data (500Hz) were collected in the frontal plane while standard video data (30Hz) were collected in the sagittal plane (see Figure 4-3). The frontal view provided the information necessary to obtain the angular displacement of the tibia relative to the femur (tibiofemoral rotation in the transverse plane). The sagittal view provided qualitative documentation of the testing procedure. A 300 ms period (starting just prior to heel strike) was digitized to determine tibiofemoral rotation for each orthotic intervention during each running trial. ASCII data from the 300 ms intervals were then processed to permit analysis of the most critical part of footstrike. The critical time periods consisted of a 140 ms span (20 ms prior to heelstrike and 120 ms after heelstrike) and were chosen based on the following rationale. Internal tibial rotation of the tibia begins with heelstrike and continues during the first 15% of the gait cycle (Wright, Desai, & Henderson, 1964). The slowest participants in the study had ground contact times less than 240 ms; therefore, half of their contact time (120 ms) easily encompassed the initial 15% of the gait cycle. Additionally, the peak heel strike force occurs between 15 and 30 ms after contact with the ground (Nigg et al., 1987) and the maximum thrust force develops between 35% and 50% of total stance time (Miller, 1990).

The other dynamic task that was studied relative to tibiofemoral rotation simulated landing from a jump and consisted of a one-legged landing after stepping off of

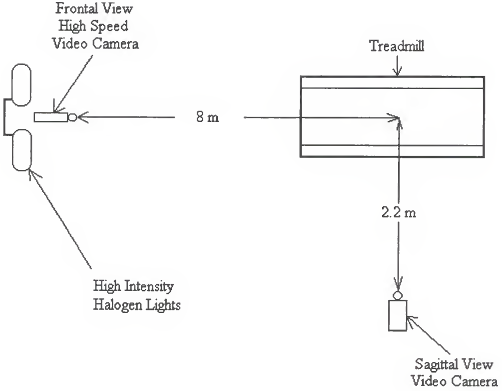


Figure 4-3. Overhead view of running experiment. A high-speed video camera was positioned in front of the treadmill to provide a frontal view of the activity. A standard video camera was positioned perpendicular to the long axis of the treadmill to provide a sagittal view of the activity.

a 40 cm high platform onto the floor. Three trials were performed by each participant with each of the four types of orthotic posting. Video data were collected in the frontal and sagittal planes. High-speed (500 Hz) digital video were used in the frontal plane to obtain information concerning tibial and femoral rotation. The digital video data were analyzed for 300 ms (starting prior to impact). Similar to the running task, ASCII data from the 300 ms intervals were narrowed to allow analysis of the most critical aspect of footstrike. Again, the important time period consisted of a 140 ms span (20 ms prior to footstrike and 120 ms after footstrike). The maximum force produced during a simple landing task occurs 20 to 40 ms after first contact (Dufek & Bates, 1991). Thus, the 120 ms period of interest included and exceeded the period necessary to capture the instant of maximum force production. The orientation of test equipment is shown in Figure 4-4.

The rotation of the tibia relative to the femur was calculated using a modification of the technique described by Cornwall and McPoil (1995). The rotation of each segment (tibia or femur) was calculated by measuring the perceived distance between the spherical reflectors on each pointing device. Because the actual distance between the markers was known it was possible to calculate the angle (α) that represents the angle defined by the relationship between the known and perceived distance between the reflective markers, as well as, the angle of rotation. The equation for the angle α follows:

$$\alpha=\cos^{-1}(X_2/X_1)$$

Where, X_1 was the actual distance between the reflective spheres and X_2 was the perceived distance between the reflective spheres (See Figure 4-5).

Values for α were calculated for the femur (α_{femur}) and the tibia (α_{tibia}). For the purpose of this project, internal rotation was considered positive and external rotation

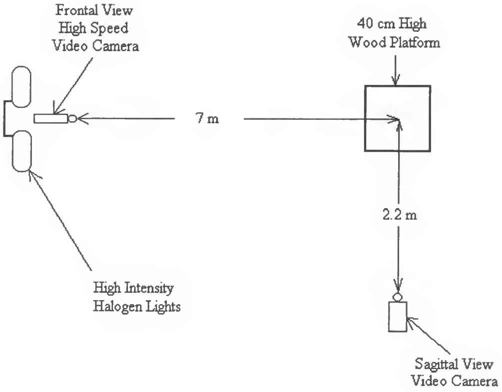


Figure 4-4. Overhead view of landing experiment. A high-speed video camera was positioned in front of the wooden platform to provide a frontal view of the activity. A standard video camera was positioned perpendicular to the direction of movement to provide a sagittal view of the activity.

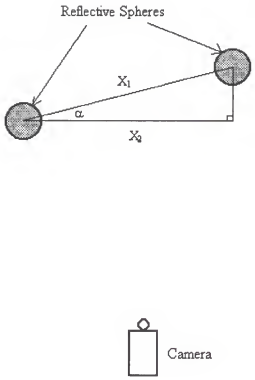


Figure 4-5. Schematic diagram displaying the technique for rotational calculations.

was negative. Therefore, it was possible to calculate the amount of relative rotation between the femur and tibia ($\alpha_{relative}$). The formula for this calculation follows:

$$\alpha_{relative} = \alpha_{tibia} - \alpha_{femur}$$

Positive values of $\alpha_{relative}$ indicated a net internal rotation of the tibia relative to the femur.

Design/Analysis

Performance means and standard deviations were calculated for each of the four dependent variables (Q-angle, thigh foot angle, subtalar joint ratio, and tibiofemoral rotation). Separate Gender x Orthotic x Trial ANOVAs (2 x 4 x 3) with repeated measures on the second and third factors were used to analyze the tibiofemoral rotation data for (a) running on a treadmill and (b) landing on a single leg after stepping off a 40 cm high platform. All statistical tests were conducted with the traditional level of significance ($\alpha = .05$). When appropriate, follow-up tests were conducted using Tukey's Honestly Significant Difference procedure or the Scheffé Post Hoc procedure.

A stepwise multiple regression equation was computed to relate the independent variable (tibiofemoral rotation) and the dependent variables (gender, activity, Q-angle, thigh foot angle, and subtalar joint ratio). The equation was as follows:

$$y = \beta_0 + \beta_1 x_1 + \beta_2 x_2 + \beta_3 x_3 + \beta_4 x_4 + \beta_5 x_5.$$

Where y is the dependent variable, β_n are constants, and x_n represent the independent variables.

Results

Participant Characteristics

Fifty-one individuals participated in the study. Participants, on average, described themselves as physically active and participating in physical activity two to three times per week. The participants studied in this experiment were a subset of those who participated in the tests performed in Study II. A summary of their physical characteristics is displayed in Table 4-1.

Tibiofemoral Rotation

Video data were analyzed for each participant. All trials were digitized automatically using PEAK5 (Peak Performance Technologies). Coordinate data points for each reflective marker (created by PEAK5) were further analyzed using Microsoft® EXCEL. Maximum internal tibiofemoral rotation values were calculated for each activity (run and jump) and orthotic posting condition (neutral, medial, lateral, and none).

Running. The mixed design ANOVA (Gender x Orthotic x Trial) revealed a significant orthotic posting main effect, $F(3, 147) = 3.25$, $p < .02$. This main effect was significant using the traditional F test, as well as, the Greenhouse-Geisser and Huynh-Feldt conservative degrees of freedom adjustments. Post-hoc analysis was performed using the Tukey's follow-up procedure and revealed that the neutral orthotics increased tibiofemoral rotation compared to the medial orthotics.

No differences in tibiofemoral rotation were detected between males and females for the running tasks or across the trials. The Gender x Orthotic x Trial ANOVA yielded no significant two or three way interactions: Posting x Gender, Trial x Gender, Posting x Trial x Gender. The results of the non significant F tests appear in Appendix G and maximum internal tibiofemoral rotation values for the running task appear in Table 4-2.

Landing. The landing task analysis revealed a significant three-way interaction Posting x Trial x Gender, $F(6, 294) = 2.47$, $p < .02$, as well as, the two-way interaction

TABLE 4-1 CHARACTERISTIC INFORMATION FOR TEST PARTICIPANTS

Characteristic	Male	Female
n	28	23
Age (yr ± SD)	21.2 ± 2.4	20.6 ± 1.0
Mass (kg ± SD)	76.3 ± 9.2	58.9 ± 9.2
Height (cm ± SD)	175.9 ± 5.0	163.0 ± 5.8

TABLE 4-2 MAXIMUM INTERNAL TIBIOFEMORAL ROTATION VALUES FOR THE RUNNING TASK

Orthotic Posting	Male	Female
None	10.3 ± 3.7°	10.3 ± 3.6°
Neutral	11.1 ± 3.8°	11.3 ± 3.2°
Medial	10.4 ± 3.2°	9.9 ± 2.8°
Lateral	10.8 ± 3.7°	10.8 ± 3.0°

Posting x Gender, $F(3, 147) = 3.14$, $p < .03$. The interactions are shown in Figures 4-6 and 4-7. Simple main effect tests were computed on the three-way interaction and Tukey's honestly significant difference procedures were used for follow-up analysis. The simple main effect tests were calculated across the three trials. Table 4-3 contains maximum internal tibiofemoral values for the jumping trials. Only the third trial differentiated the goups, $F(7, 196) = 2.33$, $p < .03$. Tukey's post-hoc procedure revealed that tibiofemoral rotation in the male no orthotic condition was greater than the tibiofemoral rotation for females in the neutral posting condition.

Concerning the two-way interaction, the Scheffé procedure was used to evaluate tibiofemoral rotation between males and females across the four orthotic conditions. The Scheffé procedure was chosen because the groups compared contained unequal numbers of data points and could not be analyzed using the Tukey procedure. No differences were detected between males and females for any of the posting conditions.

No main effect for tibiofemoral rotation was found between males and females for the landing task, $F(1, 49) = 2.82$, $p = .10$. Orthotic posting comparisons did not reveal a main effect for tibiofemoral rotation, $F(3, 147) = 1.26$, $p = .29$. No main effect for trial blocks was detected for tibiofemoral rotation, $F(2, 98) = 0.82$. The remaining tests revealed no significant interactions: Trial x Gender ($F(2, 98) = 0.06$), $p = .94$) or Posting x Trial ($F(6, 294) = 1.80$, $p = .10$). The two significant interactions were detected using traditional F tests and remained significant with the conservative degrees of freedom adjustments (Greenhouse-Geisser and Huynh-Feldt).

Prediction Equation. To create a prediction equation for tibiofemoral rotation, a stepwise linear multiple regression was calculated using a .05 level of significance for

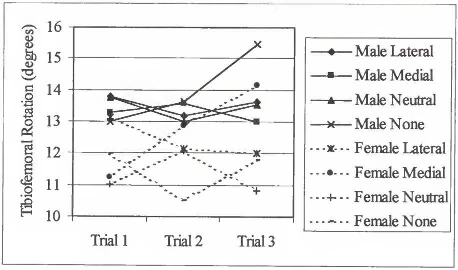


Figure 4-6. A graphical depiction of the three-way interaction Posting x Trial x Gender.

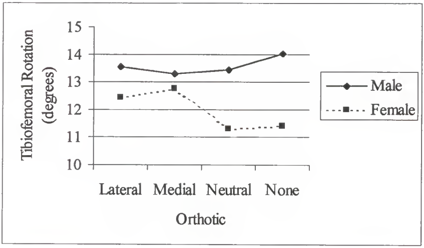


Figure 4-7. A graphical depiction of the two-way interaction Posting x Gender.

TABLE 4-3 MAXIMUM INTERNAL TIBIOFEMORAL ROTATION VALUES FOR THE ONE-LEGGED LANDING TASK

Orthotic Posting	Male	Female
None	14.0 ± 5.3°	11.4 ± 3.1°
Neutral	13.4 ± 4.8°	11.3 ± 4.2°
Medial	13.3 ± 5.0°	12.8 ± 3.9°
Lateral	13.5 ± 5.6°	12.4 ± 4.1°

adding a variable and a .10 level of significance for removing a variable that was previously added. Two data points were identified as outliers (beyond 3 standard deviations) and were removed from the analysis. Activity was the only variable retained in the prediction equation, $F(1, 98) = 17.52$, $p < .001$. Therefore, the equation took the following form:

$$y = 7.57 + 2.51x_2.$$

Where y is the maximum tibiofemoral rotation, 7.57 is the intercept, 2.51 is a coefficient, and x_2 is a variable representing the activity being performed (1 = running or 2 = jumping). The four variables excluded from the model are shown in Table 4-4. Three tests for model adequacy were performed: (a) $R^2 = 0.152$, MSE = 9.01, (b), normal probability plot, and (c) histogram of residuals. Appendix H displays the normal probability plot and histogram of residuals. The normal probability plot is nearly linear suggesting that the distribution of error terms does not deviate substantially from a normal distribution.

Discussion

Tibiofemoral rotation is a common mechanism of ACL injury (Jarvinen et al., 1994; Hess et al., 1994; Emerson, 1993). This may be due to the bony geometry of the tibiofemoral joint which dictates that excessive internal tibiofemoral rotation increases the distance between the attachment points of the ACL and induced tension in the ligament (Ahmed, Burke, Duncan, & Chan, 1992). Therefore, excessive internal rotation of the tibiofemoral joint may predispose the ACL to injury. These facts, in combination with the increased likelihood of females suffering ACL injuries provided the impetus for this study. The following hypotheses were generated for this experiment: (a) maximum

TABLE 4-4 VARIABLES EXCLUDED FROM TIBIOFEMORAL PREDICTION EQUATION

Variable	t	p
Gender	-1.51	.135
Q-angle	-1.35	.181
TFA	-0.03	.978
STJ-ratio	-0.29	.770

tibiofemoral rotation values differ between males and females, (b) maximum tibiofemoral rotation may be influenced by in-shoe orthotics, and (c) maximum tibiofemoral rotation may be predicted based on gender, the type of physical activity being performed, Q-angle, TFA, and subtalar joint ratio.

Running

During running, rotation occurs in the sagittal, frontal, and transverse planes (Hintermann & Nigg, 1998). The transverse plane movements of the knee are believed to be related to ACL injuries and were studied here. However, the normal ranges of transverse plane motion of the knee are not widely known and have not been compared between males and females.

The average of all maximum internal tibiofemoral rotations calculated for the running task was $10.6 \pm 3.4°$. Ishii and colleagues (1997) measured the relative rotation between the tibia and femur during an open-chain extension movement. Measurements indicated that the tibia was internally rotated ($10.6 \pm 2.8°$) at 60° of flexion when compared to full extension. Although the magnitudes are similar, the tasks were completely different. Zarins et al. (1983) also studied rotation of the knee in a passive manner and produced similar conclusions. For individuals with normal knees, internal rotation of the knee was limited to 10° when the knee was flexed only 5°. Again, these values are comparable to those calculated in the present work. However, Ishii et al. (1997) and Zarins et al. (1983) each used non weight bearing passive measures. Their findings described a range of possible internal tibiofemoral rotation, which include the running values computed here. Some dynamic studies have considered the relative rotation between the femur and the tibia. Lafortune et al. (1992) studied the kinematics of the knee during walking using video analysis. These authors attached reference frames to the tibia and femur of five healthy male subjects with intra-cortical pins. For the walking task, Lafortune et al. (1992) determined that the maximum internal tibiofemoral rotation was $5.0 \pm 1.6°$. Running is a more dynamic activity than walking, therefore, higher values of tibiofemoral rotation should occur during running. The data collected herein support this belief.

Interestingly, no main efffect was detected for tibiofemoral rotation between males and females. Based on the increased incidence of ACL injuries for females and the existence of tibiofemoral rotation as a mechanism of injury, it was postulated that females would experience more internal tibiofemoral rotation during running than their male counterparts. Actually, the mean value for males (10.7°) was slightly higher than the mean value for females (10.6°). This result is intriguing because the males should have been experiencing higher ground reaction forces due to their increased running velocity ($velocity_{males} = 2.32 \pm 0.32$ m/s, $velocity_{females} = 2.14 \pm 0.34$ m/s) and mass ($mass_{males} =$ 76.3 kg, $mass_{females} = 58.9$ kg). These findings indicate that gender does not affect tibiofemoral rotation. Alternatively, gender effects may have been masked by differences in running velocity and mass.

Orthotics had an influence on tibial rotation, as evidenced by the significant main effect. Unexpectedly, tibiofemoral rotation in the neutral orthotic condition ($11.2 \pm 2.7°$) was greater than tibiofemoral rotation in the medial orthotic condition ($10.2 \pm 2.0°$). Bates et al. (1979) concluded that orthotics are capable of reducing pronation during running. Therefore, orthotics should be capable of reducing internal tibial rotation and possibly internal tibiofemoral rotation. More specifically, medially posted orthotics should reduce pronation and internal tibiofemoral rotation more than the other orthotic conditions. Accordingly, the mean value for tibiofemoral rotation using the medial orthotic was the lowest of the four conditions although it was statistically significant only when compared to the neutral posting. Nawoczenski, Cook, and Saltzman (1995) reported that orthotics reduced internal tibial rotation by 2° during running for individuals that were diagnosed as pronators. The one degree difference detected in this study may have been statistically significant, but only represents a 9% decrease in rotation compared to the neutral condition and may not be biomechanically significant. Additionally, the individuals tested here were not diagnosed as pronators and thus, may have been less susceptible to the influence of the orthotics.

Landing

Jumping and subsequent landing are an integral part of sports such as basketball and volleyball (Bobbert, 1990). In deed, a task known as "drop jumping" is frequently used as a training technique for athletes that play jumping sports (Bobbert, Huijing, & van Ingen Schenau, 1987). This training technique involves dropping from a height and performing a counter-movement jump immediately upon landing. Landing involves a shock absorption phase during which the subtalar joint transfers pronation and eversion of the calcaneus into internal tibial rotation and possibly internal tibiofemoral rotation. Additionally, orthotics are commonly prescribed to control pronation and may influence the relative movement between the tibia and femur during landing from a jump. Sports that involve such movements may put individuals at risk for ACL injury and could be particularly hazardous to females. Gray et al. (1985) reported that 58% of injuries to female basketball players happen during landing from a jump and 72% of these are knee injuries. Arendt and Dick (1995) noted that female basketball players were almost four times as likely to suffer an ACL injury as their male counterparts. The problem exists, but there are no data comparing a common mechanism (tibiofemoral rotation) of ACL injury between males and females during a landing task.

The increased rate of ACL injury for females and differences in musculoskeletal alignment between males and females infers that a difference in transverse plane knee kinematics may exist between genders during dynamic tasks, such as landing from a jump. However, some results produced in this study contradict this assumption. No main effect for tibiofemoral rotation was detected between males and females. Similar to the data collected during the running portion of this study, the mean value of tibiofemoral rotation for males (13.6°) was larger than for females (12.0°). This trend is counterintuitive and cannot be attributed to the testing protocol, but could be related to the increased mass of the male participants. Jump height was not self-selected like running velocity. Both males and females executed landings from a height of 40 cm. However, the males tested had more mass than the females and most likely had correspondingly higher ground reaction forces. Based on the main effect tests, tibiofemoral rotation does not vary according to gender during a landing task and can not be solely responsible for the difference in ACL injury rates between males and females. However, gender differences are included in the significant interactions.

A three-way interaction was detected for Posting x Trial x Gender. Data were then compared across the trials. All combinations of gender and orthotic were equivalent for trial 1 and trial 2, supporting the statement that gender did not affect tibiofemoral rotation. However, males that were not wearing orthotics had significantly more internal tibiofemoral rotation (15.5°) than females wearing neutral orthotics (10.8°). Ozguven and Berme (1988) measured impact forces for gymnasts landing from a height of 0.45 m and reported maximum values ranging from 5.0 to 7.0 times body weight. For the individuals studied here and the conservative estimate of 5.0 times body weight, males would have been subjected to ground reaction forces over 3700 N and the females would have experienced forces under 2900 N. The increase in non-normalized ground reaction forces may help to explain the increased tibiofemoral rotation for males in the third trial, but does not indicate why no differences were detected for the first two trials. Increased forces should have produced higher values of tibiofemoral rotation. The increase in rotation could have been due to marker movement. The body mass index of the male participants was 24.7 kg/m^2 compared to 22.0 kg/m^2 for the female participants. A higher body mass index implies that more soft tissue could have been present on the legs of the male participants. More soft tissue could have produced additional marker movement that was not actually representative of the relative movement between the tibia and the femur. Alternatively, the increased rotation observed in males may be explained statistically. The difference may be attributable to an unexplained outlier in the male no orthotic data set (as identified by regression analysis). Without the presence of the questionable value (25.8°), the mean value for males without orthotics would decrease from 14.0° to 13.6°. The 25.8° value is physiologically possible in an open chain passive mode, but it highly unlikely in a closed chain landing.

The two-way interaction identified for Posting x Gender supports the discussion of the three-way interaction for Posting x Trial x Gender. Conservative post-hoc analyses indicated no differences between males and females in spite of the previously mentioned outlier. The lack of a significant main effect and explanations of interactions and subsequent follow-up tests provide evidence that gender had minimal influence on tibiofemoral rotation for the population studied. These findings do not support tibiofemoral rotation as the main factor that leads to ACL injury and the difference in injury rates between the genders.

As previously discussed, orthotics are often prescribed to control subtalar pronation and have been shown to influence tibial rotation. Therefore, it was hypothesized that an orthotic intervention would influence tibiofemoral rotation during landing from a jump. This hypothesis was not supported by the orthotic posting main effect test. However, the influence of orthotics was embedded in the significant interactions detected. Again, the three-way interaction detected for Posting x Trial x Gender and subsequent analysis revealed that differences were present in the third trial. Males that were not wearing orthotics had more internal tibiofemoral rotation (15.5°) than females wearing neutral orthotics (10.8°). This implies that orthotics may have had some influence on tibiofemoral rotation. The orthotics used in this study were designed to manipulate tibiofemoral rotation by inducing artificial situations of bony malalignment in the participants. The medially posted orthotics were expected to reduce tibiofemoral rotation and the laterally posted orthotics were predicted to increase tibiofemoral rotation. Perhaps, the semi-rigid orthotics may not have been stiff enough to alter the bony alignment of the foot and ankle to the degree intended or the actual movement of the bony segments may have been masked by soft tissue movement as mentioned earlier. The difference could also be attributable to the unexplained outlier (25.8°) in the male no orthotic data set. The two-way interaction identified for Posting x Gender provides more support for the discussion of the three-way interaction for Posting x Trial x Gender. Post-hoc analyses indicated no differences between males and females for any of the orthotic conditions.

The lack of a significant main effect and subsequent explanations of interactions contribute evidence that orthotics had little effect on tibiofemoral rotation for the population studied. Therefore, orthotics may not be effective in modifying tibiofemoral rotation for normal individuals performing a landing task and may not be an effective intervention relative to this proposed mechanism of ACL injury. However, the experimental population tested here may have shaped these results. The participants were not diagnosed as pronators and their pronation, tibial rotation, and tibiofemoral rotation may not have been modifiable with the addition of orthotics.

Prediction Equation

Correctly identifying individuals at risk for injury prior to the actual occurrence of injury would allow for the introduction of interventions designed to prevent tibiofemoral rotation. Regression analyses have been performed in an effort to achieve this task. Shambaugh, Klein, and Herbert (1991) created a prediction equation for risk of injury based on skeletal measures and correctly predicted injuries for 85% of 45 recreational athletes. Bennell et al. (1996) correctly predicted stress fractures in 91.7% of female track athletes, but concluded that stress fractures could not be predicted for male track atheltes. Loudon, Jenkins, and Loudon (1996) calculated a regression equation to compare ACL injured females to uninjured females. The authors concluded that excess knee recurvatum, navicular drop, and pronation were significant discriminators between the groups. However, no data were presented to indicate that the regression equation was useful for predicting future ACL injuries.

McPoil and Cornwall (1996) took another approach and used regression to predict rearfoot motion (maximum pronation). Seventeen static lower extremity measurements were made on 27 healthy young adults. Only one of the variables (navicular drop) was able to predict maximum pronation ($r^2 = 0.17$). The prediction equation for tibiofemoral rotation calculated in the present study also attempted to predict lower extremity movement based on static musculoskeletal measures, as well as, type of activity being performed and gender. The prediction equation explained an amount of variability ($r^2 = 0.15$) similar to the equation calculated by McPoil and Cornwall (1996). However, none of the musculoskeletal measures were entered into the model. Only the activity (running or landing from a jump) was able to predict the amount of tibiofemoral rotation measured for the 51 healthy participants. The landing task was associated with an increase in tibiofemoral rotation. Therefore, individuals participating in activities that require predominantly more jumping than running will incur more tibiofemoral rotation than individuals involved in running dominated tasks. Musculoskeletal alignment measures and gender were not good predictors of internal tibiofemoral rotation and may not be capable of predicting ACL injuries.

Conclusions

The bony geometry of the foot and ankle has been implicated as a risk factor for injuries to the lower limb and may be associated with a mechanism of ACL injury (internal tibiofemoral rotation). Based on this belief, the static musculoskeletal measures collected in Study II were combined with transverse plane knee kinematic information in an attempt to identify individuals predisposed to a mechanism of ACL injury. The findings suggest that gender does not influence tibiofemoral rotation during running and that orthotics have a minimal effect on tibiofemoral rotation. Results of the landing portion of the study are more convoluted although conclusions similar to those made for running can be made. Additionally, the static musculoskeletal measures collected are not capable of predicting the amount of tibiofemoral rotation experienced by the group of participants studied. Alternate static measures and/or kinematic evaluation techniques should be developed to further investigate the issue of the difference in ACL injury rates between males and females and possible interventions.

CHAPTER 5
GENDERAL DISCUSSION

Since 1972 and the introduction of Title IX, a dramatic and steady increase has occurred in female participation in athletics (Arendt & Dick, 1995). Generally, females and males are injured at similar rates if they are participating in similar sports with similar rules. However, one dramatic difference in injury rates has been observed: females suffer noncontact ACL injuries at much higher rates than males (Arendt, Agel, & Dick, 1999). This difference in ACL injury rates has been highly scrutinized, but no clear relationships have been identified between extrinsic or intrinsic factors and this form of knee injury.

The problem is important because ACL injuries are the most common severe ligamentous injuries (Wilk, Arrigo, Andrews, & Clancy, 1999) and may start a chain of events including instability, additional ligamentous injury, meniscal injury, and arthritic changes (Markolf et al., 1995). Recent improvements in surgical and rehabilitation techniques have led to better surgical outcomes. However, there is no ideal substitute for an individual's normal ACL (LaPrade & Burnett, 1994). The incidence, severity, cost, and potential for long-term disability resulting from ACL tears make their prevention a priority in the medical and research circles (Bonci, 1999). The ultimate development of interventions begins with investigations of risk factors that may contribute to injury.

Many extrinsic and intrinsic factors believed to be related to ACL injury rates have been studied. Intrinsic factors are less controllable by the individual and may prove to be the true estimators of predisposition to ACL injury. These factors have included measurements of intercondylar notch geometry, musculoskeletal alignment, and mechanisms of injury (tibiofemoral rotation during flexion and extension).

Most studies concerning factors predisposing to ACL injury and gender differences have been one-dimensional. This limitation of previous work emphasizes the need for an in-depth analysis for males and females of intrinsic factors suspected of predisposing individuals to injury, a common mechanism of injury, and the effect of in-shoe orthotics on transverse plane kinematics of the knee during dynamic activities. The present study is unique in its scope with combined skeletal alignment measures, musculoskeletal alignment measures, and transverse plane knee kinematics. Each of these elements has been examined separately, but they have not previously been evaluated in a combined manner.

Three studys were completed in the current project in an attempt to achieve the goals of (a) detecting the existence of anatomical variations in males and females believed to be related to ACL injury, (b) determining if these anatomical variations are related to a common mechanism of ACL injury, and (c) investigating an intervention designed to mimic malalignments of the foot and influence rotation of the knee during dynamic situations. A synopsis of the three studys and related research follows.

<u>Study I: Intercondylar Notch Geometry</u>

Narrow intercondylar notches have been implicated as a possible risk factor for ACL injuries. Considerable debate exists concerning notch width varying between injured and uninjured individuals, as well as, males and females. Two beliefs concerning intercondylar notch stenosis are most frequently discussed: (a) a narrow notch will have a correspondingly smaller and weaker ACL and (b) a narrow notch surrounding a normal size ACL would provide insufficient space for the ligament to function normally.

Study I of this project involved an anthropometric study of cadaver femur anatomy to determine if males and females differed concerning intercondylar notch geometry. Cadaveric skeletal data from the remains of 100 males and 100 females stored in the Terry Collection at the Museum of Natural History (Smithsonian Institution, Washington, D.C.) were studied. The distal end of each femur was digitally photographed. From these photographs, indices of notch width (NWI), notch area (NAI), and notch shape (NSI) were calculated. The individuals studied did not vary between genders when compared by NWI or NAI ($p > .017$). NSI was a statistically significant discriminator between males and females ($p < .004$). Males, on average, had larger values for NSI than females (0.638 and 0.599 respectively).

NWI. Notch width is most often reported as notch width index (NWI). Due to the close proximity between the cruciate ligaments and the intercondylar notch, the size and shape of the notch are believed to be linked to ACL injuries. Anderson et al. (1987) detected signficant differences between ACL injured individuals and normal individuals. Souryal and Freeman (1993) reported that athletes with narrow notches were more likely to suffer ACL injuries. LaPrade and Burnett (1994) concluded that intercondylar notch stenosis is associated with ACL injury, but no differences were detected between the genders. Teitz, Lind, and Sacks (1997) detected no differences between notch width indices for males and females or between normal healthy individuals and injured individuals. Curiously, NWI data are numerous but inconclusive. Additional research may help to clarify this situation.

In the present study, NWI did not vary between male and female cadaver femora. The calculated values for males (0.26) and females (0.25) were similar to those reported in previous studies. The data indicate that intercondylar notch width does not vary between males and females and therefore, can not explain the discrepancy in ACL injury rates. These findings agree with the work of Teitz, Lind, & Sacks (1997). The increased rate of ACL ruptures in females can not be attributed to differences in NWI.

NAI. Notch area index has been used sparingly as a tool for evaluating intercondylar notch geometry. Houseworth and colleagues (1987) calculated the area of the intercondylar notch and concluded that a narrow posterior arch of the intercondylar notch may be a risk factor related to ACL injury. Muneta, Takakuda, and Yamamoto (1997) concluded that the width of the intercondylar notch was not related to the size of the ACL.

Based on the data collected in Study I, the width of the intercondylar notch proved insignificant as a discriminator between males and females, but it was not the only variable related to intercondylar notch geometry analyzed. The area of the notch was also analyzed. However, NAI did not vary between males and females ($p > .05$). These data contradict the conclusions of Houseworth et al. (1987). Based on the results of this portion of Study I, the anterior cruciate ligament should pass through the same relative two-dimensional space regardless of the gender of the individual. Therefore, NAI can not be responsible for the difference in ACL injury rates between males and females.

NSI. Although relatively unstudied, a potentially important aspect of the intercondylar notch is the shape of the notch itself. Anderson et al. (1987) observed an inverted U shape for normal notches and noted narrow notches to be more waveshaped. No objective studies have been formed to quantify the shape of the intercondylar notch.

The third variable of intercondylar notch geometry analyzed in this project dealt with the relative shape of the notch. NSI was greater for males than females. Thus, the shape of the notch in males, on average, may be more round than in females. A lower NSI could be detrimental to normal function of the ACL during certain situations, such as, when the knee is in full extension and the cruciate ligaments are pulled tight and reside in the more anterior portion of the intercondylar notch. A low NSI indicates that this particular region of the intercondylar notch will be more narrow and will provide less space for the ligament to function correctly. A notch that is relatively wider and more round provides more space for the ACL when the knee is near or in full extension. This difference in shape of the intercondylar notch may contribute to the difference in injury rates between males and females.

Study II: Musculoskeletal Alignment

The second part of this project focused on anthropometric measures of living individuals. Physically active college-aged adults were recruited as participants. Measures of Q-angle, thigh foot angle, and subtalar joint motion were collected from the participants. Each of these factors are believed to influence tibial rotation and possibly tibiofemoral rotation. Thirty healthy males and twenty-seven healthy females were measured and Q-angle in females was found to be greater than in males ($p < .017$). TFA and subtalar joint ratio did not vary between males and females ($p > .017$), although the

Bonferonni t test for TFA ($p = .019$) approached the adjusted level of statistical significance (.017) required for multiple comparisons.

Q-angle. Another anatomical factor that has been implicated as predisposing to ACL injuries is Q-angle. Large Q-angles have been associated with lower limb injuries and knee extensor dysfunction (Cowan et al., 1996; Ciullo, 1993; Hunter, 1984). Theoretically, a greater Q-angle increases the horizontal component of the quadriceps muscle force and pulls the patella laterally and may render the ligaments and menisci of the knee vulnerable to injury (Hsu et al., 1990). The majority of the Q-angle research has indicated that females have larger Q-angles than males (Hsu et al., 1990; Woodlang & Francis, 1992). Conversely, Livingston and Mandigo (1997) detected no significant differences in Q-angle between males (10.5°) and females (12.2°).

The Q-angle is an estimate of the direction of pull of the quadriceps muscle (Schulthies et al., 1995) and has been implicated in several disorders of the knee (Horton & Hall, 1989). Specifically, high values appear to be detrimental to knee function. The Q-angle values collected in Study II are similar to previously reported data and add to the body of knowledge indicating that females have larger Q-angles than males.

A larger Q-angle may result in greater subtalar joint pronation (Subotnick, 1981). Greater subtalar joint pronation may cause excessive internal rotation of the tibia and increase the distance between the attachment points of the ACL producing tension in the ligament (Ahmed, Burke, Duncan, & Chan, 1992). Therefore, excessive internal rotation of the tibiofemoral joint partially resulting from a large Q-angle may predispose the ACL to injury. The larger Q-angle values recorded for the females studied may help to explain the difference in ACL injury rates between the genders.

Thigh Foot Angle. Thigh foot angle is a combined measure of tibiofibular torsion and the alignment of the foot (Stuberg et al., 1991). The bony alignment of the lower extremity, including high values of TFA, has been implicated as predisposing to overuse injuries (Krivickas, 1997) and ACL rupture (Heugel, 1999). Excessive internal tibiofemoral rotation may also result from excessive external tibial torsion, measured as TFA. Only a single study has attempted to link TFA to ACL injury and no research has been performed comparing TFA between genders or in relation to a mechanism of injury.

TFA data presented here contradict data from the few available studies that have measured TFA, but the difference is most likely due to differences between the populations studied. No significant differences were detected between normal males and females concerning TFA although there is strong trend ($p = .019$) suggesting that females may have larger values of TFA than their male counterparts. However, based on the data collected in Study II, TFA can not be conclusively linked to the difference in ACL injury rates between males and females.

Subtalar Joint Axis. The bony geometry of the foot and ankle has been implicated as a risk factor for injuries to the lower limb. Specifically, excessive foot eversion and abnormal tibial rotation have been associated with knee injuries (Hintermann & Nigg, 1993). The relationship between these two movements is determined by the subtalar joint axis. Manter (1941) determined the location of the axis on cadaver lower limbs and reported that the mean axis lies 42° above the horizontal plane and 16° medial to the midline of the foot (see Figure 1-5). Individuals with high subtalar joint axes may be more at risk of ACL injury upon compensatory pronation of the subtalar joint than those with lower subtalar joint axes because they have a greater amount of internal tibial rotation with a given amount of subtalar joint pronation (see Figure 1-6).

Subtalar joint pronation, a combination of eversion, abduction, and dorsiflexion, compensates for a variety of lower extremity malalignments (Kitaoka, Lundberg, Luo, & An, 1995). The internal rotation of the tibia occurring concurrently with subtalar joint pronation affects knee kinematics and may induce tension in the ACL. Despite its apparent importance, no clinical methods have been developed to calculate the location of the subtalar joint axis.

An alternative to actually locating the subtalar joint axis may be to determine the net effect of the axis. This can be accomplished by calculating a ratio of tibial rotation to calcaneal movement. The computed subtalar joint ratios did not vary between genders for the participants in this study. This indicates that individuals have much greater tibial rotation for any amount of calcaneal movement than previously hypothesized. This finding may be particularly important relative to the prior discussion of a common mechanism of ACL injury (internal tibiofemoral rotation). A high subtalar joint ratio may result in a greater amount of internal tibial rotation for any amount of calcaneal eversion. Therefore, based on data collected in Study II, the subtalar joint ratio may partially explain the difference in ACL injury rates between genders. Additionally, a high ratio may be a clinically measurable risk factor for any individual regardless of gender.

Additionally, the close relationship between tibial rotation, subtalar joint pronation, and malalignments of the foot suggests that management of the compensatory motions through foot orthotic intervention may control excessive motions and reduce the risk of knee injury. Foot orthotics are commonly used to control subtalar joint movement and may be used to mimic abnormal foot alignments. Chiumento (1998) conducted research to identify the link between tibiofemoral rotation and orthotic interventions for female participants and determined that orthotics influenced tibiofemoral rotation during landing from a jump. The possibility of orthotics altering transverse knee kinematics is intriguing and deserving of additional research.

Study III: Mechanisms of Injury

The majority of ACL injuries occur in a noncontact fashion (Noyes et al., 1983). In a survey of ACL injured skiers, 88% of the injuries occurred in the presence of tibiofemoral rotation (Jarvinen et al., 1994). The predominant injury mechanisms for the female skiers studied were knee flexion and internal rotation (Jarvinen et al., 1994). Hess and colleagues (1994) observed a similar trend with the most injuries resulting from a combination of flexion and internal tibial rotation. Emerson (1993) noted that change of direction when the tibia is internally or externally rotated can cause ACL injury. Tibial rotation appears to be the common link. Despite the importance of understanding mechanisms of injury, only Chiumento (1998) has studied tibiofemoral rotation during dynamic activities.

The third study of this project was a kinematic analysis of knee mechanics during two dynamic activities. Specifically, tibiofemoral rotation was analyzed using high speed video. Twenty-eight males and twenty-three females from Study II participated in Study III. Each participant performed a series of drop jump landings and ran on a treadmill. In-shoe orthotics were introduced in an attempt to manipulate tibiofemoral rotation as a result of simulating foot malalignments during the dynamic activities. The anthropometric data collected in Study II were combined with the kinematic data of

Study III to explore the existence of relationships between static musculoskeletal measures suspected of predisposing an individual to ACL injury and dynamic measures of a mechanism of ACL injury (tibiofemoral rotation). No differences in tibiofemoral rotation were detected between males and females for the running task. Differences were detected relative to orthotic posting condition for the running task ($p < .02$) although the difference was only 1° and is most likely not biomechanically significant. Main effects tests indicated that tibiofemoral rotation did not vary between males and females ($p > .05$) and orthotic posting had no effect for the landing task relative to tibiofemoral rotation ($p > .05$). However, gender by orthotic effects were identified in two significant interactions. Orthotic intervention may affect males and females differently when compared by tibiofemoral rotation. Alternatively, the measured differences may reflect the presence of statistical outliers.

Running. During running, rotation occurs in the sagittal, frontal, and transverse planes (Hintermann & Nigg, 1998). The transverse plane movements of the knee are believed to be related to ACL injuries and were studied here. However, the normal ranges of transverse plane motion of the knee are not widely known and have not been compared between males and females. The internal tibiofemoral rotation data collected here correspond to passive and active values reported in previous research. Interestingly, no differences in tibiofemoral rotation were detected between males and females. Based on the increased incidence of ACL injuries for females and the existence of tibiofemoral rotation as a mechanism of injury, a notion was created that females would experience more internal tibiofemoral rotation during running than males. These data indicate that gender does not affect tibiofemoral rotation.

Orthotics did have an influence on the measured amount of tibial rotation. Unexpectedly, tibiofemoral rotation in the neutral orthotic condition was greater than in the medial orthotic condition. Medially posted orthotics should reduce pronation and internal tibiofemoral rotation more than the other orthotic conditions, but the mean value for tibiofemoral rotation using the medial orthotic was the lowest of the four conditions. However, it was statistically significant only when compared to the neutral posting. The findings indicate that the mechanism of injury studied (tibiofemoral rotation during running) does not vary based on gender and that orthotics influence tibiofemoral rotation during running for healthy males and females, but most likely to a level that is not biomechanically significant. Therefore, based on the treadmill running data, tibiofemoral rotation during running does not help to explain the discrepancy in ACL injury rates in females.

Landing. Jumping and subsequent landing are an integral part of sports such as basketball and volleyball (Bobbert, 1990). Landing involves a shock absorption phase during which the subtalar joint transfers pronation and eversion of the calcaneus into internal tibial rotation and possibly internal tibiofemoral rotation. Sports that involve such movements may put individuals at risk for ACL injury and appear to be particularly hazardous to females based on their higher injury rates.

The increased rate of ACL injury for females and differences in musculoskeletal alignment between males and females infers that a difference in transverse plane knee kinematics may exist between genders during dynamic tasks, such as landing from a jump. However, some findings collected in this study contradict this assumption and the work of Chiumento (1998). No main effect for tibiofemoral rotation was detected between males and females, however, unique combinations with gender were revealed in significant interactions.

The Posting x Trial x Gender interaction was detected and differences were revealed in the third trial. Specifically, males that were not wearing orthotics had significantly more internal tibiofemoral rotation (15.5°) than females wearing neutral orthotics (10.8°). The difference could be attributable to increased ground reaction forces for males because of additional body weight, increased body mass index for males causing added marker movement, but most likely is due to the presence of an unexplained outlier.

A second interaction, Posting x Gender, supports the discussion of the above three-way interaction. Follow-up analyses indicated no differences between males and females in spite of the presence of an outlier. The lack of a significant main effect and explanations of interactions and subsequent follow-up tests provide evidence that gender had minimal influence on tibiofemoral rotation for the population studied. These findings do not explain the difference in injury rates between the genders.

As previously discussed, orthotics are often prescribed to control subtalar pronation and have been shown to influence tibial rotation. Therefore, it was hypothesized that an orthotic intervention would influence tibiofemoral rotation during landing from a jump. This hypothesis was not supported by the orthotic posting main effect test. However, the influence of orthotics was embedded in the significant interactions detected. All combinations of gender and orthotic were equivalent for trial 1 and trial 2 suggesting that level of orthotic intervention did not affect tibiofemoral rotation. However, males that were not wearing orthotics had more internal tibiofemoral rotation (15.5°) than females wearing neutral orthotics (10.8°). This implies that orthotics may have had some influence on tibiofemoral rotation. The unexpected manner in which the orthotic effect appeared may be attributable to the type of orthotic tested (semi-rigid) or the previously mentioned outlier. The two-way interaction identified for Posting x Gender provides more support for the discussion of the three-way interaction for Posting x Trial x Gender because no main effect differences between males and females for any of the orthotic conditions were detected.

The lack of a significant main effect and subsequent explanations of interactions contribute evidence that indicates that the orthotics had little effect on tibiofemoral rotation for the population studied. Therefore, based on the current data, orthotics may not be effective in modifying tibiofemoral rotation for normal individuals performing a landing task and may not be an effective intervention for a mechanism of ACL injury. However, the test population measured in this experiment may have shaped these results because the individuals were not diagnosed with foot malalignments that would normally require orthotic intervention. Additionally, only the kinematics of the knee were studied. The orthotic interventions and physical tasks may have altered movements at other segments in the kinetic chain (foot, ankle, or hip) without affecting the transverse plane kinematics of the knee.

Prediction Equation. The ultimate goal of research related to ACL injury is to successfully predict which individuals are at risk and introduce an intervention to try and prevent injury. Regression analyses have been performed in an effort to achieve this task. The prediction equation for tibiofemoral rotation calculated in the present study attempted to predict lower extremity movement based on static musculoskeletal measures, the type of activity being performed, and gender. The prediction equation explained only a small amount of the total variability (15%) although this value is similar to the 17% reported by McPoil and Cornwall (1996). Only the activity (running or landing from a jump) was able to predict the amount of tibiofemoral rotation measured for the 51 healthy participants. The landing task was associated with an increase (2.5°) in tibiofemoral rotation. Therefore, individuals participating in activities that require a great deal of jumping will incur more tibiofemoral rotation than individuals involved in running dominated tasks. Musculoskeletal alignment measures and gender proved to be poor predictors of internal tibiofemoral rotation and may not be suitable for use in predicting ACL injuries.

Summary and Conclusions

The current findings contribute to the body of knowledge related to factors predisposing individuals to mechanisms of ACL injury and gender differences (see Appendix I for a listing of the findings of each study). This study was the first of its kind to investigate factors possibly related to ACL injury in such a comprehensive manner. The cause of ACL injuries is most likely a multifactorial problem and therefore should be addressed in a comprehensive fashion. The combination of results from the three studys that made up this project help to confirm previously held theories concerning factors possibly predisposing to ACL injury and dispel others. Additionally, the findings point to further avenues of research.

Traditional measures of intercondylar notch geometry do not vary between males and females without history of ACL injury. However, a new measure of notch shape (NSI) does distinguish between males and females. This index may help to explain the higher rate of ACL injury observed in females. Several malalignments of the musculoskeletal system have also been implicated as risk factors that predispose individuals to ACL injury and were hypothesized to differ between males and females. Q-angle in females was larger than in males and may partially explain the difference in injury rates between the genders. Based on the belief that the bony geometry of the foot and ankle may be related to tibiofemoral rotation, static musculoskeletal measures were combined with transverse plane knee kinematic information in an attempt to identify individuals predisposed to a mechanism of ACL injury. Findings indicated that gender had a minimal influence on tibiofemoral rotation, tibiofemoral rotation is relatively unaltered by an orthotic intervention, and tibiofemoral rotation can not be predicted by measures of static musculoskeletal alignment. The static measures studied (Q-angle, thigh foot angle, and subtalar joint ratio) do not dictate the functional movements of the knee.

This project represented an initial step in reducing ACL injury by identifying risk factors and specifically addressed gender issues in a comprehensive fashion. Measurable factors that can be directly linked to ACL injury must be identified. Solutions must be found to reduce the risk of injury for individuals prone to injury as indicated by predictive measureable factors. Finally, preventative interventions must be tested and optimized to bring the problem of ACL injury under control.

REFERENCES

Ahmed, A. M., Burke, D. L., Duncan, N. A., & Chan, K.H. (1992). Ligament tension pattern in the flexed knee in combined passive anterior translation and axial rotation. Journal of Orthopaedic Research, 10, 854-867.

Agletti, P., Insall, J. N., & Cerulli, G. (1983). Patellar pain and incongruence. I: Measurements of incongruence. Clinical Orthopedics, 176, 217-224.

Anderson, A. F., Lipscomb, A. B., Liudahl, K. J., & Addlesone, R. B. (1987). Analysis of the intercondylar notch by computed tomography. American Journal of Sports Medicine, 15(6), 547-552.

Arendt, E. A. (1994). Orthopedic issues for active and athletic women. Clinics in Sports Medicine, 13, 486-495.

Arendt, E. A., Agel, J., & Dick, R. (1999). Anterior cruciate ligament injury patterns among collegiate men and women. Journal of Athletic Training, 34(2), 86-92.

Arendt, E. A., & Dick, R. (1995). Knee injury patterns among men and women in collegiate basketball and soccer. American Journal of Sports Medicine, 23, 694-701.

Bass, W. M. (1987). Human osteology: A laboratory and field manual. (3rd ed.). Columbia, MO.

Bates, B. T., Osternig, L. R., Mason, B., & James, L. S. (1979). Foot orthotic devices to modify selected aspects of lower extremity mechanics. American Journal of Sports Medicine, 7(6), 338-342.

Bennell, K. L., Malcolm, S. A., Thomas, S. A., Reid, S. J., Brukner, P. D., Ebeling, P. R., & Wark, J. D. (1996). Risk factors for stress fractures in track and field athletes. American Journal of Sports Medicine, 24(6), 810-817.

Bobbert, M. F. (1990). Drop jumping as a training method for jumping ability. Sports Medicine, 9(1), 7-22.

Bobbert, M. F., Huijing, P. A., & van Ingen Schenau, G. J. (1987). Drop jumping I. The influence of jumping technique on the biomechanics of jumping. Medicine and Science in Sports and Exercise, 19(4), 332-338.

Bonci, C. M. (1999). Assessment and evaluation of predisposing factors to anterior cruciate ligament injury. Journal of Athletic Training, 34(2), 155-164.

Cappozzo, A., Cappello, A., Della Croce, U., & Pensalfini, F. (1997). Surface-marker cluster design criteria for 3-d bone movement reconstruction. IEEE Transactions of Biomedical Engineering, 44(12), 1165-1174.

Cheng, J. C., Chan, P. S. Chiang, S. C., & Hui, P. W. (1991). Angular and rotational profile of the lower limb in 2,630 Chinese children. Journal of Pediatric Orthopaedics, 11, 154-161.

Chiumento, A. (1998). Can transverse kinematics of the female knee be altered by functional foot orthotics during dynamic tasks? Unpublished master's thesis, University of Florida, Gainesville.

Ciullo, J. V. (1993). Lower extremity injuries. In: Pearl, A. J. (Ed.), The athletic female. (pp.267-298). Champaign, IL: Human Kinetics.

Cornwall, M. W., & McPoil, T. G. (1995). Footwear and foot orthotic effectiveness research: A new approach. Journal of Orthopedic and Sports Physical Therapy, 21, 337-344.

Cowan, D. N., Jones, B. H., Frykman, P. N., Polly, D. W. Jr., Harman, E. A., Rosenstein, R. M., & Rosenstein, M. T. (1996). Lower limb morphology and risk of overuse injury among male infantry trainees. Medicine and Science in Sports and Exercise, 28, 945-952.

Dufek, J. S. & Bates, B. T. (1991). Biomechanical factors associated with injury during landing in jump sports. Sports Medicine, 12(5), 326-337.

Emerson, R. J. (1993). Basketball knee injuries and the anterior cruciate ligament. Clinics in Sports Medicine, 12(2), 317-327.

Feagin, J. S., Cabaud, H. E., & Curl, W. W. (1982). The anterior cruciate ligament: Radiologic and clinical signs of successful and unsuccessful repairs. Clinical Orthopedics, 164, 54-58.

Feagin, J. A., Lambert, K. L., Cunningham, P. R., Anderson, L. M., Riegel, J., King, P. H., & VanGenderen, L. (1987). Consideration of anterior cruciate ligaments injury in skiing. Clinical Orthopaedics and Related Research, 216, 3-18.

Goris, J. E. & Graf, B. K. (1996). Risk factors for anterior cruciate ligament injury. Wisconsin Medical Journal, 367-369.

Gray, J., Taunton, J. E., McKenzie, D. C., Clement, D. B., McConkey, J. P., & Davidson, R. G. (1985). A survey of injuries to the anterior cruciate ligament of the knee in female basketball players. International Journal of Sports Medicine, 6, 314-316.

Griffis, N. O., Vequist, S. W., & Yearout, K. M. (1989). Injury prevention of the ACL. American Orthopedic Society for Sports Medicine. Traverse City, MI.

Heiple, K. G. & Lovejoy, C. O. (1971). The distal anatomy of australopithecus. American Journal of Physical Anthropology, 35(1), 75-84.

Hess, T., Rupp, S., Hopf, T., Gleitz, M., & Liebler, J. (1994). Lateral tibial avulsion fractures and disruptions to the anterior cruciate ligament: A clinical study of their incidence and correlation. Clinical Orthopaedics and Related Research. 303, 193-197.

Heugel, M., Meister, K., Rolle, G., Indelicato, P. A., & Horodyski, M. B. (1999). Rotational alignment of the lower extremity as a risk factor for noncontact injury to the ACL in the female athlete. Manuscript submitted for publication.

Hicks, J. H. (1953). The mechanics of the foot. I. The joints. Journal of Anatomy, 87, 345-357.

Hintermann, B. & Nigg, B. M. (1993). Pronation from the viewpoint of the transfer of movement between the calcaneus and the tibia. Schweiz Z Sportmed, 41(4), 151-156.

Horton, M. G. & Hall, T. L. (1989). Quadriceps femoris muscle angle: Normal values and relationships with gender and selected skeletal measures. Physical Therapy, 69, 897-901.

Houseworth, S. W., Mauro, V. J., & Mellon, B. A. (1987).The intercondylar notch in acute tears of the ACL: A computer graphics study. American Journal of Sports Medicine, 15, 221-224.

Hsu, R. W. W., Himeno, S., Coventry, M. B., & Chao, E. Y. S. (1990). Normal axial alignment of the lower extremity and load-bearing distribution of the knee. Orthopaedics and Related Clinical Research, 255, 215-227.

Hunter, L. Y. (1984). Women's athletics: The orthopedic surgeon's viewpoint. Clinics in Sports Medicine, 3(4), 809-827.

Huston, L. J. & Wojtys, E. M. (1996). Neuromuscular performance characteristics in elite female athletes. American Journal of Sports Medicine 24(4), 427-436.

Hutchinson, M. R. & Ireland, M. L. (1995). Knee injuries in female athletes. Sports Medicine, 19, 288-302.

Hvid, I. Andersen, L. I., & Schmidt, H. (1981). Chondromalacia patellae: The relation to abnormal patellofemoral joint mechanics. Acta Orthopedica Scandanavia, 52, 661-666.

Inman, V. T. (1981). Human walking. Baltimore: Williams and Wilkins.

Inman, V. T. (1976). The joints of the ankle. Baltimore: Williams and Wilkins.

Irrgang, J. J., Safran, M. R., & Fu, F. H. (1996). The knee: ligamentous and meniscal injuries. In J. E. Zachazewski, D. J. Magee, & W. S. Quillen (Eds.), Athletic injuries and rehabilitation (pp.623-632). Philadelphia, PA: Saunders.

Ishii, Y., Terajima, K., Terashima, S., & Koga, Y. (1997). Three-dimensional kinematics of the human knee with intracortical pin fixation. Clinical Orthopedics, 343, 144-150.

Jarvinen, M., Natri, A., Laurila, S., & Kannus, P. (1994). Mechanisms of anterior cruciate ligament ruptures in skiing. Knee Surgery, Sports Traumatology, Arthroscopy, 2(4), 224-228.

Kernozek, T. W. & Greer, N. L. (1993). Quadriceps angle and rearfoot motion: Relationships in walking. Archives of Physical Medicine and Rehabilitation, 74, 407-410.

Kitaoka, H. B., Lundberg, A., Luo, Z. P., & An, K. (1995). Kinematics of the normal arch of the foot and ankle under physiologic loading. Foot & Ankle International, 16(8), 1995.

Krivickas, L. S. (1997). Anatomical factors associated with overuse sports injuries. Sports Medicine, 24(2), 132-146.

Krogman, W. M. (1968). The human skeleton in forensic medicine. Springfield, IL: C.C. Thomas.

Lafortune, M. A., Cavanagh, P. R., Sommer, H. J., & Kalenak, A. (1992). Three-dimensional kinematics of the human knee during walking. Journal of Biomechanics, 25(4), 347-357.

LaPrade, R. F., & Burnett, Q. M. (1994). Femoral intercondylar notch stenosis and correlation to anterior cruciate ligament injuries: A prospective study. American Journal of Sports Medicine, 22(2), 198-202.

Livingston, L. A. (1998). The quadriceps angle: A review of the literature. Journal of Orthopedic and Sports Physical Therapy, 28(2), 105-109.

Livingston, L. A., & Mandigo, J. L. (1997). Bilateral within-subject Q angle asymmetry in young adult females and males. Biomedical and Scientific Instrumentation, 22, 112-117.

Loudin, J. K., Jenkins, W., & Loudin, K. L. (1996). The relationship between static posture and ACL injury in female athletes. Journal of Orthopaedic and Sports Physical Therapy, 24, 91-97.

Lovejoy, O. C. & Heiple, K. G. (1970). A reconstruction of the femur of australopithecus africanus. American Journal of Physical Anthropology, 32(1), 33-40.

Malone, T. R., Hardaker, W. T., Garrett, W. E., Feagin J. A., & Bassett, F. H. (1993). Relationship of gender to anterior cruciate ligament injuries in intercollegiate basketball players. Journal of Southern Orthopaedic Association 2, 36-39.

Manter, J. T. (1941). Movements of the subtalar and transverse tarsal joints. The Anatomical Record, 80(4), 397-410.

Marieb, E. N. & Mallatt, J. (1992). Human anatomy. Redwood City, CA. Benjamin/Cummings Publishing Company, Inc.

Markolf, K. L, Burchfield, D. M., Shapiro, M. M., Shepard, M. F., Finerman, G. A. M., & Slauterbeck, J. L. (1995). Combined knee loading states that generate high anterior cruciate ligament forces. Journal of Orthopaedic Research 13, 930-935.

McClay, I. & Bray, J. (1996). The subtalar angle: A proposed measure of rearfoot structure. Foot & Ankle International, 17(8), 499-502.

McCulloch, M. U., Brunt, D., & Vander Linden, D. (1993). The effect of foot orthotics and gait velocity on lower limb kinematics and temporal events of stance. Journal of Orthopaedic and Sports Physical Therapy, 17, 2-10.

McPoil, T. G. & Cornwall, M. W. (1996). The relationship between static lower extremity measurements and rearfoot motion during walking. Journal of Orthopedic and Sports Physical Therapy, 24(5), 309-314.

Miller, D. I. (1990). Ground reaction forces in distance running. In P. R. Cavanagh (ed.), Biomechanics of distance running (pp. 203-224). Champaign, IL: Human Kinetics.

Muneta, T., Takakuda, K., & Yamamoto, H. (1997). Intercondylar notch width and its relation to the configuration and cross-sectional area of the anterior cruciate ligament. A cadaveric knee study. American Journal of Sports Medicine, 25(1), 69-72.

Nawocezenski, D. A., Cook, T. M., & Saltzman, C. L. (1995). The effect of foot orthotics on three-dimensional kinematics of the leg and rearfoot during running. Journal of Orthopaedic and Sports Physical Therapy, 21, 317-327.

Nigg, B. M., Bahlsen, H. A., Luethi, S. M., & Stokes, S. (1987). The influence of running velocity and midsole hardness on external impact forces in heel-toe running. Journal of Biomechanics, 20(10), 951-959.

Norwood, L. A., Jr. & Cross, M. (1977). The intercondylar shelf and the anterior cruciate ligament. American Journal of Sports Medicine, 5, 171-176.

Novak, P. J., Bach, B. R. Jr., Bush-Joseph, C. A., & Badrinath, S. (1996). Cost containment: A charge comparison of the anterior cruciate ligament reconstruction. Arthroscopy, 12(2), 160-164.

Noyes, F. R., Matthews, A. M., & Butler, D. S. (1983). The symptomatic ACL deficient knee. Journal of Bone and Joint Surgery, 65, 154-174.

Ozguven, H. N. & Berme, N. (1988). An experimental and analytical study of impact forces during human jumping. Journal of Biomechanics, 21(12), 1061-1066.

Paulos, L. E. (1992). Why failures occur symposium: Revision ACL surgery. Proceedings of the American Orthopaedic Society for Sports Medicine Eighteenth Annual Meeting, San Diego.

PEAK Performance Technologies Inc. (1992). PEAK V User's Reference Manual. Engelwood, CO.

Pearl, A.J. (1993). The athletic female. Champaign, IL: Human Kinetics.

Phillips, R. D. & Lidtke, R. H. (1992). Clinical determination of the linear equation for the subtalar joint axis. Journal of the American Podiatric Medical Association, 82(1), 1-20.

Reinschmidt, C., van den Bogert, A. J., Nigg, B. M., Lundberg, A., & Murphy, N. (1997). Effect of skin movement on the analysis of skeletal knee joint motion during running. Journal of Biomechanics, 30(7), 729-732.

Robson, M. (1903). Ruptured crucial ligaments and their repair by operation. Annals of Surgery, 37, 716.

Rockar, P. A. (1995). The subtalar joint: Anatomy and joint motion. Journal of Orthopedic and Sports Physical Therapy, 21(6), 361-372.

Root, M. L, Orien, W. P., & Weed, J.H. (1977). Clinical Biomechanics-Vol. II. Los Angeles: Clinical Biomechanics.

Rosenberg, K. R. (1992). The evolution of human childbirth. Yearbook of Physical Anthropology. Suppl. 15:89-124.

Schierman, G. L. & Cheetham, P. J. (1990). Motion measurement using the PEAK Performance Technologies motion measurement system. Society of Photo-optical Instrumentation and Engineering Procedures, 1356, 63-70.

Schulthies, S. S., Francis, R. S., Fisher, A. G., & Van De Graaff, K. M. (1995). Does the Q angle reflect the force on the patella in the frontal plane? Physical Therapy, 75(1), 24-30.

Shambaugh, J. P., Klein, A., & Herbert, J. H. (1991). Structural measures as predictors of injury in basketball players. Medicine and Science in Sports and Exercise, 23(5), 522-527.

Shelbourne, K. D., Davis, T. J., & Klootwyk, T. E. (1998). The relationship between intercondylar notch width of the femur and the incidence of anterior cruciate ligament tears. A prospective study. American Journal of Sports Medicine, 26(3), 402-408.

Souryal, T. O. & Freeman, T. R. (1993). Intercondylar notch size and anterior cruciate ligament injuries in athletes. A prospective study. American Journal of Sports Medicine, 21(5), 723.

Souryal, T. O., Moore, H. A., & Evans, J. P. (1988). Bilaterality in ACL injuries: Associated intercondylar notch stenosis. American Journal of Sports Medicine, 16, 449-454.

Stewart, T. D. (1979). Essentials of forensic anthropology. Springfield, IL: C.C. Thomas.

Stuberg, W, Temme, J., Kaplan, P., Clarke, A., & Fuchs, R. (1991). Measurement of tibial torsion and thigh-foot angle using goniometry and computed tomography. Clinical Orthopedics, 272, 208-212.

Subotnick, S. I. (1975). Orthotic foot control and the overuse syndrome. Physician and Sportsmedicine, 3, 32-38.

Tanzer, M. & Lenczner, E. (1990). The relationsihp of intercondylar notch size and content to notchplasty requirement in anterior cruciate ligament surgery. Arthroscopy, 6(2), 89-93.

Teitz, C. C., Lind, B. K., & Sacks, B. M. (1997). Symmetry of the femoral notch width index. American Journal of Sports Medicine, 25(5), 687-690.

Terry G. C. (1980). The anatomy of the extensor mechanism. Clinics in Sports Medicine, 8(2), 163-177.

Tiberio, D. (1987). The effect of excessive subtalar joint pronation on patellofemoral mechanics: A theoretical model. Journal of Orthopaedic and Sports Physical Therapy, 9(4), 160-164.

Timm, K. E. (1997). The clinical and cost-effectiveness of two different programs for rehabilitation following ACL reconstruction. Journal of Orthopedic and Sports Physical Therapy, 25(1), 43-48.

Tortora, G. J. (1999). Principles of human anatomy (7th ed.). New York, NY: Harper Collins College Publishers.

van den Bogert, A. J., Smith, G. D., & Nigg, B. M. (1994). Invivo determination of the anatomical axes of the ankle joint complex: An optimization approach. Journal of Biomechanics, 27(12), 1477-1488.

Weaver, D. S. (1980). Sex differences in the ilia of a known sex and age sample of fetal and infant skeletons. American Journal of Physical Anthropology, 52, 191-195.

Wilk, K. E., Arrigo, C., Andrews, J. R., & Clancy, W. G. (1999). Rehabilitation after anterior cruciate ligament reconstruction in the female athlete. Journal of Athletic Training, 34(2), 177-193.

Woodlang, L. H. & Francis, R. S. (1992). Parameters and comparisons of the quadriceps angle of college-aged men and women in the supine and standing positions. American Journal of Sports Medicine, 20(2), 208-211.

Wright, D. G., Desai, S. M., & Henderson, W. H. (1964). Action of the subtalar and ankle-joint complex during the stance phase of walking. Journal of Bone and Joint Surgery, 46-A(2), 361-382.

Zarins, B., Rowe, C. R., Harris, B. A., & Watkins, M. P. (1983). Rotational motion of the knee. American Journal of Sports Medicine, 11(3), 152-156.

APPENDIX A
EQUIPMENT SPECIFICATIONS

Equipment for Study I

Model	Manufacturer	Location
Olympus D-200L Digital Camera	Olympus America, Inc.	Melville, NY
Toshiba 445CDT Computer	Toshiba America, Inc.	New York, NY
Digital Camera Software (ver 3.0)	Olympus America, Inc.	Melville, NY
MATLAB (ver 5.2)	Mathworks, Inc.	Natick, MA
Image Processing Toolbox (ver 2.1)	Mathworks, Inc.	Natick, MA
SPSS (ver 7.5)	SPSS, Inc.	Chicago, IL

Equipment for Study II

Model	Manufacturer	Location
Goniometer (#FE054)	Micro Bio-Medics, Inc.	Mount Vernon, NY
Angle Finder	Dasco Pro, Inc.	Rockford, IL
Laser Pointer (LX 200)	Tandy Corporation	Houston, TX
Toshiba 445CDT Computer	Toshiba America, Inc.	New York, NY
SPSS (ver 7.5)	SPSS, Inc.	Chicago, IL

Equipment for Study III

Model	Manufacturer	Location
Goniometer (#FE054)	Micro Bio-Medics, Inc.	Mount Vernon, NY
Angle Finder	Dasco Pro, Inc.	Rockford, IL
Laser Pointer (LX 200)	Tandy Corporation	Houston, TX
ProForm Crosswalk GT treadmill	ICON Health & Fitness, Inc.	Logan, UT
Kodak Motion Corder Analyzer (#SR-500)	Eastman Kodak Corporation	San Diego, CA
Halogen Lamps (#TSQ1000)	Regent Lighting Corporation	Burlington, NC
JVC Video Camera (#GR-AX730)	JVC Company of America	Wayne, NJ
PEAK5 (ver 5.1.1)	Peak Performance Technologies, Inc.	Englewood, CO
Brooks Paragon Athletic Shoes	Brooks Sports, Inc.	Bothell, WA
Spenco ¾ Length Arch Supports	Spenco Medical Corporation	Waco, TX
4° High Density Polyethylene Posting Material	UCO International	Prospect Heights, IL
Toshiba 445CDT Computer	Toshiba America, Inc.	New York, NY
SPSS (ver 7.5)	SPSS, Inc.	Chicago, IL

APPENDIX B
MATLAB SUBROUTINE FOR ANALYZING INTERCONDYLAR NOTCH GEOMETRY

```
% input filename
R=input('Which image should be analyzed?');
y=int2str(R);
v= '-1.jpg';
t = strcat(y,v);

% open data file
fid=fopen('wmale.dat','a');
I=imread(t);
figure(1)
imagesc(I);

% measure block width
disp('While maintaining a horizontal line, left click on the left side of the cube')
disp('right click on the right side of the cube')

X=getline;
A=round(sqrt(sum( (X(1,:)-X(2,:)).^2)));
disp('The width of the cube is ')
disp(A)
disp('pixels')


% measure notch width
disp('While maintaining a horizontal line, left click on the left side of the notch (at the
widest point)')
disp('right click on right side of notch')
X=getline;
B=round(sqrt(sum( (X(1,:)-X(2,:)).^2)));
disp('The width of the notch is ')
disp(B)
disp('pixels')


% measure condyle width
disp('While maintaining a horizontal line, left click on the left side of the femoral condyle
(at level of notch measurement)')
disp('right click on right side of femoral condyle')
X=getline;
C=round(sqrt(sum( (X(1,:)-X(2,:)).^2)));
disp('The width of the femoral condyles is ')
disp(C)
disp('pixels')


% calculate notch width index
```

```
NWI=B/C

% measure notch height
disp('While maintaining a vertical line, left click on the highest point of the notch')
disp('right click on bottom side of the cube')
X=getline;
D=round(sqrt(sum( (X(1,:)-X(2,:)).^2)));
disp('The height of the notch is ')
disp(D)
disp('pixels')


% calculate notch shape index
NSI=B/D

% measure notch area
disp('Left click on the lower left portion of the notch, level to the base of cube')
disp('continue to left click on the edge of the notch using a minimum of 10 points')
disp('the last click (a right click) should be on the lower right portion of the notch, level
to the base of cube')
p1 = roipoly;
area1 = bwarea(p1);
E=round(area1);
disp('The area of the notch is ')
disp(E)
disp('square pixels')

% measure condyle area
disp('Left click on the lower left portion of the notch, level to the base of cube')
disp('continue to left click on the edge of the notch using a minimum of 10 points')
disp('continue tracing the outer edge of the condyle from right to left using a minimum of
25 points')
disp('the last click (a right click) should be horizontal and left of the first click')
p2 = roipoly;
area2 = bwarea(p2);
F=round(area2)
disp('The area of the condyles is ')
disp(F)
disp('square pixels')

%calculate notch area index
NAI=E/F
RESULTS = [ R A B C D E F NWI NSI NAI ]
fprintf(fid,'%g %g %g %g %g %g %g %g %g %g \n',RESULTS)
status = fclose(fid);
```

APPENDIX C
PHYSICAL ACTIVITY QUESTIONNAIRE

Physical Activity Questionnaire: Please read, fill in the blank, or circle the most appropriate response.

Participant Number __________

1. In regard to physical activity, how would you describe your life?

   Very active / Active/ Average / Not very active / Inactive

2. Do you exercise regularly to keep physically fit? (not including organized sports).

   Yes / No

3. Over the last month, how often did you exercise or play sports for 15 minutes or more?

   No exercise or sports in last month / Less than once per week / One time per week / Two or three times per week / Four or more times per week

4. In the last month, how many times did you run or jog more than 15 minutes actual running time?

   None / Less than 1 time per week / About 1 time per week /

   2 to 3 times per week / 4 or more times per week

5. In the last month, did you do any vigorous exercises or sports other than running?

   Yes / No

If yes, what exercises or sports?

____

And, how many times per week ___

6. Are you right or left-handed?

   Right / Left / Both

7. Which foot do you prefer to kick a ball with?

   Right / Left / Both

APPENDIX D
INSTITUTIONAL REVIEW BOARD LETTER OF APPROVAL

UNIVERSITY OF
FLORIDA

Institutional Review Board

98A Psychology Bldg.
PO Box 112250
Gainesville, FL 32611-2250
(352) 392-0433
Fax (352) 392-0433

DATE:	08-Apr-98
TO:	Mr. Mark Tillman and Ms. Kendra Smith 151B FLG
FROM:	C. Michael Levy, Chair University of Florida Institutional Review Board
SUBJECT:	**Approval of Project # 1998 - 306**
TITLE:	Gender Defferences and Predisposition Towards Anterior Cruciate Ligament Injuries : A Biomechanical and Antropological Approach.
FUNDING:	Opportunity Fund and the National Athlet

I am pleased to advise you that the University of Florida Institutional Review Board has recommended approval of this project. Based on its review of your protocol, the UFIRB determined that this research presents no more than minimal risk to participants. Given your protocol, It is essential that you obtain signed documentation of informed consent from each participant. Enclosed is the dated, IRB-approved informed consent to be used when recruiting participants for the research.

If you wish to make any changes to this protocol, you must disclose your plans before you implement them so that the Board can assess their impact on your project. In addition, you must report to the Board any unexpected complications arising from the project that affect your participants.

If you have not completed this project by 08-Apr-99, please telephone our office (392-0433) and we will tell you how to obtain a renewal.

It is important that you keep your Department Chair informed about the status of this research project.

CML/h2

Cc: Vice President for Research
Dr. Jeffrey Bauer

APPENDIX E
INFORMED CONSENT AGREEMENT

Informed Consent Agreement

Project Title: Gender Differences and Predisposition Towards Anterior Cruciate Ligament Injuries: A Biomechanical and Anthropological Approach

Please read this consent agreement carefully before you decide to participate in this study.

Purpose of the study:
The purposes of this investigation are to examine the differences in leg anatomy between males and females and the effects of different leg anatomy and in-shoe orthotics on knee movements.

What you will do in the study:

- You will fill out a questionnaire. You will be fitted with black spandex pants. You will be wearing shoes provided by the investigator. You will be weighed and measured (height, hip width, and several other similar measurements). You will sit on the floor and reach towards your toes and as far beyond as possible. You will be fitted with reflective markers on your hips and legs. Your hips and legs will be photographed. You will jog on a treadmill at a comfortable pace for 5 minutes. You will be fitted with an in-shoe orthotic. You will jog on the treadmill for 3 more minutes while being videotaped. You will repeat this three more times with different shoe and orthotic combinations. You will jog a total of 17 minutes. You will step off of a 40cm high platform three times in each of the 4 shoe and orthotic combinations. Your hips and legs will be videotaped during the 12 landings. Your face will not be visible in the video data and ,therefore, you will not be recognizable. After the data are analyzed the tapes will be erased.

Time required:
45 minutes

Risks:
There is no greater risk involved in participating in this study than in jogging or jumping. If you have any physical limitations preventing you from walking or jumping you will be excluded from the study.

Benefits/Compensation:
There is neither compensation nor other direct benefits to you for participation.

Confidentiality:
The information you give will be handled confidentially to the extent provided by law. Your information will be assigned a code number. The list connecting your name to this number will be kept in a locked file. Video tapes and photographs will be kept in a locked cabinet. Only researchers directly involved with the project will have access to the video tapes and photographs. When the study is completed and the data have been analyzed, the list will be destroyed. Your name will not be used in any report.

Voluntary Participation:
Your participation in the study is completely voluntary. There is no penalty for not participating.

Right to withdraw from the study:
You have the right to withdraw from the study at anytime without penalty.

Whom to contact if you have questions about the study:
Mark D. Tillman, M.S., Graduate Student, Dept. of Exercise and Sport Sciences, 151B FLG, (352) 392-0580 x 234
Kendra R. Smith, M.F.S., Graduate Student, Dept. of Anthropology ,1350 Turlington Hall, (352) 392-6772
Jeffrey A. Bauer, Ph.D., Assistant Professor, Dept. of Exercise and Sport Sciences, 151A FLG, (352) 392-0584 x 263
Anthony B. Falsetti, Ph.D., Assistant Professor, Dept. of Anthropology, 1350 Turlington Hall, (352) 392-6772

Whom to contact about your rights in the study:
UFIRB Office, Box 112250, University of Florida, Gainesville, FL 32611-2250

Agreement:
I have read the procedure described above. I voluntarily agree to participate in the procedure and I have received a copy of this description.

Participant: ______________________________ **Date:** ____________
Principal Investigator: ______________________________ **Date:** ____________

APPENDIX F

INSTITUTIONAL REVIEW BOARD FORM

UNIVERSITY OF FLORIDA INSTITUTIONAL REVIEW BOARD

COMPLETE THIS FORM USING A WORD PROCESSOR OR TYPEWRITER. Before completing read reverse side of form.

1. **TITLE OF PROJECT:**
Gender Differences and Predisposition Towards Anterior Cruciate Ligament Injuries: A Biomechanical and Anthropological Approach

2. **PRINCIPAL INVESTIGATOR(s):** *(Name, degree, title, dept., address, phone #, e-mail & fax)*
- Mark D. Tillman, M.S., Graduate Student, Dept. of Exercise and Sport Sciences 151B FLG, 392-0580 ext. 234, tillman@nervm.nerdc.ufl.edu
- Kendra R. Smith, M.F.S., Graduate Student, Dept. of Anthropology ,1350 Turlington Hall, 392-6772 krs2@ufl.edu

3. **SUPERVISOR** (IF PI IS STUDENT): *(Name, campus address, phone #, e-mail & fax)*
- Jeffrey A. Bauer, Ph.D., Assistant Professor, Dept. of Exercise and Sports Science, 151A FLG, 392-0584 ext. 263, jbauer@hhp.ufl.edu
- Anthony B. Falsetti, Ph.D., Assistant Professor, Dept. of Anthropology, 1350 Turlington Hall, 392-6772, falsett@nersp.nerdc.ufl.edu

4. **DATES OF PROPOSED PROJECT**: From ____9/98_____ To ____9/99_____

5. **SOURCE OF FUNDING FOR THE PROJECT:** Submitted to the Opportunity Fund and the National Athletic Trainers' Association
(As indicated to the Office of Research, Technology and Graduate Education)

6. **SCIENTIFIC PURPOSE OF THE INVESTIGATION:**
The primary purposes of this investigation are to examine the differences in leg anatomy between males and females and the effects of different leg anatomy and in-shoe orthotics on knee movements. Specifically, we will examine the rotation of the knee during jogging and landing from a jump using video analysis.

7. **DESCRIBE THE RESEARCH METHODOLOGY IN *NON-TECHNICAL* LANGUAGE:**
The UFIRB needs to know what will be done with or to the research participant(s).
- We plan to enroll 70 college-age individuals that are physically active. The participant will fill out a questionnaire. The participant will be fitted with black spandex pants. A series of anthropometric measurements (height, weight, hip width, etc.) will be made on the participant. The participant will sit on the floor and reach forward as far as possible. Reflective markers will be placed on the hips and legs of the participant. The legs and hips of the participant will be photographed. The participant will be fitted with shoes. The participant will warm-up by jogging on a treadmill for 5 minutes at a comfortable pace. Following the warm-up, the participant will rest for 5 minutes. The participant will jog on a treadmill for 3 more minutes at the same comfortable pace while being videotaped. This will be repeated four times for the different conditions of orthotic (none, neutral, medial posting, and lateral posting). The total jogging time will be 17 minutes with a short rest between trials to switch in-shoe orthotics. The participant will step off of a 40cm high platform while being videotaped. Three trials will be performed with each orthotic. A total 12 landings will be performed. The participant's face will not be photographed or video taped, therefore, they will not be recognizable. Only researchers directly involved with the project will have access to the video tapes and photographs. The video tapes will be kept in a locked cabinet until all data are analyzed. After the data are analyzed, the tapes will be erased.

8. **POTENTIAL BENEFITS AND ANTICIPATED RISK:**
(If risk of physical, psychological or economic harm may be involved, describe the steps taken to protect participant.)
There is no greater risk involved in participating in this study than in a normal jogging or jumping situation.

9. **DESCRIBE HOW PARTICIPANT(S) WILL BE RECRUITED, THE NUMBER AND AGE OF THE PARTICIPANTS, AND PROPOSED COMPENSATION (if any):**
Seventy volunteers will be recruited from undergraduate and graduate classes at the University of Florida. There is no compensation for this study.

10. **DESCRIBE THE INFORMED CONSENT PROCESS. INCLUDE A COPY OF THE INFORMED CONSENT**
DOCUMENT (if applicable).
See attached Consent Form.

Please use attachments ONLY when space on the form is insufficient.

Principal Investigator's Signature	Supervisor's Signature
Principal Investigator's Signature	Supervisor's Signature
	Department Chair's Signature

APPENDIX G
NON SIGNIFICANT STATISTICAL RESULTS FOR THE RUNNING TASK

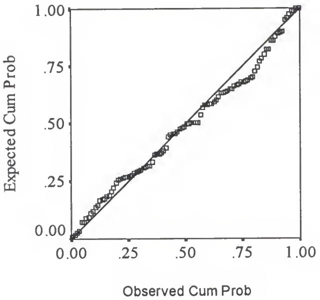
NON SIGNIFICANT STATISTICAL RESULTS FOR THE RUNNING TASK

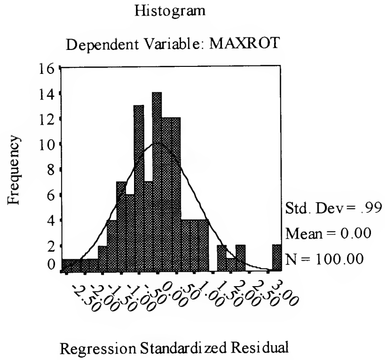
Test	F-values
Gender Main Effect	$F(1, 49) = 0.02$
Trial Block Main Effect	$F(2, 98) = 2.28$
Posting x Gender	$F(3, 147) = 0.29$
Trial Block x Gender Interaction	$F(2, 98) = 1.57$
Posting x Trial Block x Gender Interaction	$F(6, 294) = 0.58$

APPENDIX H
NORMAL PROBABILITY PLOT FOR TIBIOFEMORAL PREDICTION EQUATION

Normal P-P Plot of Regression Standardized Residual

Dependent Variable: MAXROT




Histogram
Dependent Variable: MAXROT
Frequency
16
14
12
10
8
6
4
2
0
Std. Dev = .99
Mean = 0.00
N = 100.00
-2.50
-2.00
-1.50
-1.00
-.50
0.00
.50
1.00
1.50
2.00
2.50
3.00
Regression Standardized Residual

APPENDIX I
SUMMARY OF VARIABLES MEASURED AND STATISTICAL RESULTS

SUMMARY OF VARIABLES MEASURED AND STATISTICAL RESULTS

Study I: Do males and females differ in intercondylar notch geometry?

Variable	Male	Female	p
NWI	0.255 ± 0.028	0.247 ± 0.032	.160
NAI	0.173 ± 0.022	0.170 ± 0.025	.372
NSI	0.638 ± 0.089	0.599 ± 0.094	.003*

* = Significant difference ($p \leq .05$) between male and female.

Study II: Do males and females differ in musculoskeletal alignment?

Variable	Male	Female	p
Q-angle (° ± SD)	13.1 ± 3.0	17.5 ± 3.8	.001★
TFA (° ± SD)	8.8 ± 5.2	12.4 ± 5.9	.020
Subtalar joint ratio ± SD	2.35 ± 1.7	2.10 ± 1.6	.573

★ = Significant difference ($p \leq .017$) between male and female.

Study III: A comparison of knee kinematics between males and females.

Running

Source	df	SS	MS	F	p
Gender	1	0.7388	0.7388	.02	.898
Error	49	2178.8	44.5		
Posting	3	103.7	34.6	3.25	.024*
Posting x Gender	3	9.2	3.1	0.29	.834
Error	147	1561.7	10.6		
Trials	2	29.7	14.8	2.28	.107
Trials x Gender	2	20.4	10.2	1.57	.214
Error	98	637.0	6.5		
Posting x Trials	6	47.2	7.9	0.89	.499
Posting x Trials x Gender	6	30.8	5.1	0.58	.744
Error	294	2583.7	8.8		

* Neutral orthotics (11.2°) were associated with increased tibiofemoral rotation compared to the medial orthotics (10.2°, $p \leq .05$).

Orthotic Posting	Male	Female
None	10.3 ± 3.7°	10.3 ± 3.6°
Neutral	11.1 ± 3.8°	11.3 ± 3.2°
Medial	10.4 ± 3.2°	9.9 ± 2.8°
Lateral	10.8 ± 3.7°	10.8 ± 3.0°

Landing

Source	df	SS	MS	F	p
Gender	1	386.5	386.5	2.82	.099
Error	49	6708.6	136.9		
Posting	3	40.7	13.6	1.26	.290
Posting x Gender	3	101.5	33.8	3.14	.027*
Error	147	1582.8	10.8		
Trials	2	23.3	11.6	0.82	.443
Trials x Gender	2	1.6	0.8	0.06	.944
Error	98	1389.8	14.2		
Posting x Trials	6	108.1	18.0	1.80	.098
Posting x Trials x Gender	6	147.9	24.7	2.47	.024*
Error	294	2937.5	10.0		

* Tibiofemoral rotation in the male no orthotic condition (15.5°) was greater than the tibiofemoral rotation for females in the neutral posting condition (10.8°) in the third trial ($p \leq .05$).

Orthotic Posting	Male	Female
None	14.0 ± 5.3°	11.4 ± 3.1°
Neutral	13.4 ± 4.8°	11.3 ± 4.2°
Medial	13.3 ± 5.0°	12.8 ± 3.9°
Lateral	13.5 ± 5.6°	12.4 ± 4.1°

Prediction Equation

Model

$y = 7.57 + 2.51x_2$

Variable

Y (tibiofemoral rotation)

7.57 (intercept)

2.51 (coefficient)

x_2 (activity)

* = Activity is a significant predictor of tibiofemoral rotation ($p \leq .05$).

BIOGRAPHICAL SKETCH

Mark D. Tillman was born July 2, 1970, in Ft. Lee, Virginia. After relocating several times during childhood, he moved to Crystal River, Florida, in 1981. Following high school graduation, he enrolled at the University of Florida.

During his education at the University of Florida, Mark became interested in Sports medicine and obtained bachelor's and master's degrees in Engineering Sciences and Engineering Mechanics, specializing in Biomedical Engineering. A quest for interesting coursework led him to the Department of Exercise and Sport Sciences and eventually to doctoral work. A combination of new academic information and a history of knee injuries provided the foundation for his dissertation.

Mark will continue to be involved with higher education. He has accepted a position as Assistant Professor at Stetson University in DeLand, Florida. Mark plans to continue with his line of research and will be teaching a variety of courses in Sport and Exercise Science.

I certify that I have read this study and that in my opinion it conforms to acceptable standards of scholarly presentation and is fully adequate, in scope and quality, as a dissertation for the degree of Doctor of Philosophy.

James H. Cauraugh, Chairman
Associate Professor of Exercise and Sport Sciences

I certify that I have read this study and that in my opinion it conforms to acceptable standards of scholarly presentation and is fully adequate, in scope and quality, as a dissertation for the degree of Doctor of Philosophy.

Jeffrey A. Bauer, Co-chairman
Assistant Professor of Exercise and Sport Sciences

I certify that I have read this study and that in my opinion it conforms to acceptable standards of scholarly presentation and is fully adequate, in scope and quality, as a dissertation for the degree of Doctor of Philosophy.

Mark H. Trimble
Assistant Professor of Physical Therapy

I certify that I have read this study and that in my opinion it conforms to acceptable standards of scholarly presentation and is fully adequate, in scope and quality, as a dissertation for the degree of Doctor of Philosophy.

Edward K. Walsh
Professor of Aerospace Engineering, Mechanics, and Engineering Science

This dissertation was submitted to the Graduate Faculty of the College of Health and Human Performance and to the Graduate School and was accepted as partial fulfillment of the requirements for the degree of Doctor of Philosophy.

August, 1999

Dean, College of Health and Human Performance

Dean, Graduate School